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THE UNIVERSITY OF ALBERTA

Design, Implementation, and Calibration of a Specialized
X-ray CT Scanner for Bone Evaluation

by

Derek Martin Wells

A THESIS

SUBMITTED TO THE FACULTY OF GRADUATE STUDIES AND RESEARCH
IN PARTIAL FULFILMENT OF THE REQUIREMENTS FOR THE DEGREE
OF Master of Science

Department of Applied Sciences in Medicine

EDMONTON, ALBERTA

Spring 1986

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The undersigned certify that they have read, and recommend to the Faculty of Graduate Studies and Research, for acceptance, a thesis entitled Design, Implementation, and Calibration of a Specialized X-ray CT Scanner for Bone Evaluation submitted by Derek Martin Wells in partial fulfilment of the requirements for the degree of Master of Science.

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Abstract

A specialized CT scanner for bone density evaluation has been developed. The high spatial resolution and precision of the scanner should allow improvements in diagnosis and monitoring of trabecular bone loss in the appendicular skeleton. The radiation source is a small X-ray tube, normally operated at 50 kVp with a copper filter of 0.70 mm thickness, and with a useable 50° fan beam angle. A range of object sizes between 75 mm and 220 mm diameter can be scanned, and a variable source/center-of-rotation (SCR) distance allows small objects to be "magnified" in the image reconstruction plane, thus improving relative geometrical resolution. The photon detection system consists of 26 NaI(Tl) crystal-photomultiplier units which operate in the pulse counting mode.

Data are collected in the rotate-only mode, but with 26 detector units, multiple rotations with varying detector offsets are necessary to gain the required spatial resolution. By changing the detector offset for each rotation, and with 5 rotations for a "scan", a spatial resolution of 0.6 mm can be realized which is comparable to or better than the spatial resolution of typical commercial whole body CT scanners.

The unique data collection scheme which is used for this specialized CT scanner can introduce new types of image reconstruction artifacts; these include the effects caused by inaccurate geometrical factors such as SCR distance,

fan-beam centering, or initial fan-offset. Various calibration procedures have been implemented to minimize these errors and to correct for additional factors such as dead-time, background count, and beam-hardening errors. The precision and tolerance levels (with respect to producing artifacts in the reconstructed images) for each of these calibration procedures have been established. In addition, the stability of the system has been monitored over time to determine how often re-calibration procedures are required.

Acknowledgements

I would like to express my gratitude to my supervisor, Dr. T.N. Hangartner, for his willingness to share his expertise in CT, and for the moral support he has given. This project involved the efforts of a number of people whom I would like to recognize: Mr. N. Ouellette (for the design and construction of various electronic components); Mr. E. Haska and Mr. B. Heath (for the design, testing, and implementation of the microprocessor and VAX programs); Ms. H. Vanderheide (for her advice on various practical aspects of CT); Mr. I. Yamamoto and Mr. D. Heinz (for the design and construction of various mechanical components); and Dr. T.R. Overton (for acting as my supervisor in Dr. Hangartner's absence and for his review of this manuscript). Lastly, I would like to thank all of my friends in the Department of Applied Sciences in Medicine who have made this experience most enjoyable.

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List of Abbreviations

ADFR	Activate, depress, free, and repeat
ARF	Activate, resorb, and form
ΔB	Bone mass per unit volume
BMD	Bone mineral density
BRU	Bone remodelling unit
CAMAC	Computer automated measurement and control
CR	Center-of-rotation
CRU	Communication register. unit
CT	Computed tomography
DECT	Dual energy computed tomography
DIAC	Data input accepted
DIPR	Data input ready
DOAC	Data output accepted
DOPR	Data output ready
EEGE	Exponential edge gradient effect
EPROM	Erasable programmable read-only memory
FWHM	Full width half maximum
IGOR	Input gate output register
I/O	Input/output
LED	Light emitting diode
N_o	Observed count-rate
N_t	True count-rate
μP	Microprocessor
PC	Program counter
PMT	Photomultiplier tube
PSC	Power supply controller
PSI	Programmable systems interface
PTH	Parathyroid hormone
RAM	Random access memory
ROI	Region of interest
SCR	Source/center-of-rotation
SD	Standard deviation
SECT	Single energy computed tomography
SINT	# steps before interrupt sent to scalers (interval size)
SPEED	Motor frequency
STPCNT	Number of steps issued to motor
STSNM	Steps to start next motor
T_d	Dead-time
T/P	Tolerance/precision
VAX	Virtual address extension
V_o	Common discriminator threshold voltage
WP	Workspace pointer
$\bar{x}$	Mean value of variable x

1. Introduction

Computed tomography' (CT) is the reconstruction, by computer, of a tomographic plane (slice) of an object based on measurements of transmitted radiation. In the radiological field, it is created from multiple X-ray attenuation measurements made at various angles through the object. The quality of the reconstructed image depends upon: the nature of the radiation source and the detector units; the number and speed of the individual measurements; the details of the reconstruction algorithm; the machine characteristics, e.g. precision and accuracy of the scanner movements; the methods of data display and interpretation.

The spatial reconstruction of the internal structure of an object from multiple projections is not a new concept. As early as 1917 mathematical studies relevant to the theory of image reconstruction were conducted by Radon [1]. During the late 1950s and 1960s image reconstruction techniques were being used for such diverse applications as mapping regions of the sun which emit microwave radiation (Bracewell, [2]) and the study of biomolecular structure by electron microscopy (DeRosier and Klug, [3]).

Early studies of image reconstruction for radiological applications were carried out independently by Oldendorf in 1961 [4], Kuhl and Edwards in 1963 [5], and Cormack in 1963 [6]. Cormack developed a mathematically accurate method of reconstructing images from X-ray projections; however his -----
'from the Greek words *tomos* meaning cut and *graphein* meaning to draw or to paint

work stimulated very little interest in the radiological community at that time primarily because available computers were not of sufficient size and speed to be of any practical use. The development of the first CT equipment to show clinical potential was accomplished by Hounsfield in 1967 [7]. Initial images reconstructed using his machine clearly showed a tumor in a preserved specimen of a diseased human brain; the scan took 9 hours to complete.

Full-scale design and development of a practical CT scanner for head measurements began in 1970 and the first production machine was installed at the Atkinson Morley's Hospital, Wimbledon, in 1971. Following the demonstration of this new equipment and the first published accounts of its use [8], the interest in CT as a new diagnostic tool grew rapidly. The accomplishments of Cormack and Hounsfield in the development of CT were recognized with the award, jointly, of the 1979 Nobel Prize in Physiology or Medicine.

Today there are essentially five classes of CT scanners. The type of machine which Cormack and Hounsfield developed are termed "first generation" scanners. These scanners have 1 photon source and 1 radiation detector rigidly fixed with respect to one another. Data are collected as the source/detector assembly translates across the object; a small rotation about the object at the machine center occurs between successive translations. This translate-rotate process is repeated until projection views are collected over 180° . Today, first generation scanners

are obsolete for clinical studies because of their slow scan speed (about 4 to 5 minutes per slice).

Second generation scanners collect data in the same manner as first generation scanners except that more detectors (typically 2-10) are used. For each translation of the source-detector assembly, multiple views are obtained, so that scan times are reduced to less than two minutes.

Third generation scanners have a single photon source and many radiation detectors (typically 300-700) mounted rigidly with respect to one another. Data are collected in a rotate-only mode, that is, no translational movements are required. The maximum object diameter which can be scanned is limited by the size of the fan beam, and the sampling resolution is fixed by the distance between detectors in the assembly. However, scan times are very fast (5-10 seconds), and such machines now have a wide distribution.

Fourth and fifth generation scanners further reduce scan times by eliminating the need to rotate the bulky detector assembly of the third generation systems. Fourth generation scanners consist of a fixed detector ring over 360° with only the source moving. Fifth generation scanners have both detectors and source(s) stationary. Such machines are very complex and have high costs; however, scan times are extremely fast (fractions of seconds). For example, it is possible to monitor the pumping action of the human heart with some of these systems [9].

1.1 Bone Evaluation Using CT

A major area of new interest for CT applications is the evaluation of bone density *in vivo*. Unlike the usual qualitative aspects of CT images, such as displaying gross anatomical structures, determination of bone density is a quantitative procedure where accuracy and precision of measurements are critical. Due to wide variation in "normal" bone density levels among the adult population, even when age and sex are taken into consideration, a single bone density measurement has no real significance. Instead, serial studies to determine how bone density changes over time are necessary. For this reason, the precision of the bone evaluation method is of greater importance than its accuracy. The precision of bone density measurements depends on three factors:

1. The ability to scan the same skeletal "volume" for each measurement in a serial study. Bone density at a macroscopic level varies greatly throughout the skeleton, and even throughout each measured bone section. Precise positioning of a patient is usually hard to accomplish, and thus poor precision in bone density measurements will occur if only one measurement is made. It is now common practice to take a number of scans at a particular skeletal site and get an average density value over a small volume of bone (about 4-15 cm³) to minimize such positioning errors. The limitation to this method is that multiple scans result in increased scanning time which

increases patient discomfort and in turn may give rise to movement errors.

2. Maximizing the number of picture elements (pixels) in the region for which the bone density values are estimated.

In order to improve precision, bone density values are averaged over a region of the bone (generally the inner 40 to 60% area of a section). Improving the relative geometrical resolution, i.e. maximizing the number of beam rays that pass through the object, increases the number of pixels within the region of interest. More data points result in better average values for bone density, and thus precision is ultimately improved.

3. Reducing the relative error in the counting statistics.

An X-ray transmission projection value of N counts has a relative error of $\sqrt{N}/N$. Hence by increasing the number of counts recorded, this error can be reduced and the precision of the method improved, but only at the cost of increased radiation dose to the patient.

Any bone evaluation method must take these three factors into consideration.

The CT method is particularly suited to monitoring the effects of the bone disease "osteoporosis". Unlike the diseases "rickets" and "osteomalacia", where the chemical composition of the bone changes, osteoporosis is simply a structural change in bone for which no biochemical analysis of blood or urine can define the severity of the disease. Details of the "new" bone biology and of methods to monitor

bone density changes with specific reference to the disease osteoporosis are given in Appendix I. Osteoporosis is characterized by a bone porosity such that spontaneous, non-traumatic fractures can occur. These fractures are predominantly seen in three regions: the vertebral column, the femoral neck, and the wrist. The ideal site for monitoring osteoporosis is the vertebral column because the earliest effects (vertebral fracture) of decreased bone mass are seen in this region. Whole-body commercial CT scanners have been shown to be quite sufficient for precise bone mineral density (BMD) evaluation in serial studies of this region [10]. However, one problem associated with scanning the vertebral column is that it comprises a small cross-sectional area of the torso, so that the geometrical resolution of the bone in the reconstructed image is not optimal.

Monitoring trabecular bone density in peripheral regions of the skeleton, such as the wrist, does have certain advantages. For example, smaller, cheaper scanners can be used, and problems associated with geometrical resolution can be reduced. Several investigators [11,12] have shown that there appears to be a high correlation between BMD changes in the wrist and the spine, implying that wrist measurements may be used to predict the extent of the disease osteoporosis in the spinal region of a patient. In addition, wrist measurements can be used to monitor various bone treatment protocols.

Whole body scanners have limited or no SCR adjustment; so small objects such as the wrist are not "magnified" in the reconstruction plane very well. Third generation commercial CT systems are available with SCR movement, but like any commercial system, the ability of the user to adjust various parameters of the hardware (e.g. scanner movement) and software (e.g. convolution kernel) of the system is restricted. Hence, "homemade" CT systems where the user has total control over these hardware and software parameters do have their advantages in a research environment.

A second generation scanner, specifically developed for peripheral bone density measurements, was built at the University of Zurich in 1974 [13], and a further refined model was constructed at the University of Alberta in 1979 [14]. The U of A scanner uses an ^{125}I point source and 8 NaI(Tl)/PMT detector units. Changes in bone density can be determined with a precision of 0.5% and an average skin dose per scan of less than 20 mrad. However, these CT systems, with their long scan times (about 2 minutes), can lead to motion artifacts in the reconstructed images. In addition, positioning errors are more likely to occur because fewer scans can be performed during each patient session.

1.2 Design Specifications of a New CT Scanner

What is needed is a CT system which combines good geometrical resolution for small object sizes, low dose per scan, and fast scan times, while still collecting enough information per projection to reduce statistical noise in the reconstructed image to "acceptable" levels. Faster scan times would enable more scans to be performed in one patient session with fewer motion artifacts, thereby improving the precision of the bone density measurement. A new CT scanner with these capabilities would allow monitoring of smaller changes in bone density, and earlier diagnosis of bone disease.

Key design specifications of the new CT scanner would be: (1) Data collection in a rotate-only mode so that fast scan times are possible; (2) a good relative geometrical resolution for small and large objects, i.e. the source/center-of-rotation distance or "magnification" must be variable; (3) a high spatial resolution (comparable to commercial scanners); (4) low dose so that serial studies can be routinely performed - this would involve using a highly efficient counting system; (5) a low cost, easy to maintain system which would feature: a small number of detectors, an X-ray tube instead of a radioisotope source, and a small gantry system, i.e. specialized for measuring bone density in appendicular sites of the human skeleton; and (6) a low photon energy to improve contrast resolution for bone density determination.

This thesis will describe the design, implementation, and calibration of a new generation CT scanner with the above features. The hardware and software design of the system were done by T.N. Hangartner [15] and are described in chapters 2 and 3. Chapter 4 includes the various calibration procedures which were implemented such as:

- (1) matching the X-ray spectra of each of the detector units;
- (2) setting an appropriate discriminator threshold level;
- (3) dead-time, background count, and beam-hardening corrections;
- (4) determining geometrical parameters such as SCR distance, detector angles, and centering of fan beam;
- and (5) determining the thermal characteristics of the anode.

Chapter 5 describes the unique data collection scheme of this scanner, and chapter 6 shows some of the results of the new scanner: for example, a comparison of spatial resolution between a number of CT systems is done, and the causes of certain reconstruction artifacts are explained with reference to the various calibration procedures of chapter 4. The precision and tolerance levels (with respect to producing artifacts in the reconstructed images) for each of the calibration procedures are also determined, and the stability of the system is monitored over time to determine how often re-calibration procedures are required. Future developments of the CT system such as new detector packaging schemes and dual energy CT are briefly mentioned in chapter 7.

2. Hardware

This section will describe the major components of the CT unit (Figure 1). These include : (1) mechanical features of the gantry system, (2) control rack which houses the microprocessor, (3) detection system, (4) X-ray generator, and (5) VAX 11/750 computer and CAMAC interface system. Special emphasis will be placed on systems which are unique to this CT unit.

2.1 Gantry System

The scanner is capable of translational movements in three directions as well as a rotational movement. In the "home" or reset position of the scanner (Figure 2) the source/detector assembly is positioned for the beginning of a scan. The height of the front support is adjusted to position the object (a plexiglass phantom in this photo) at the center-of-rotation (CR) of the scanner.

2.1.1 Front Plate

Movement of the front plate is necessary for centering the fan beam with respect to the CR of the scanner. This plate, which supports the X-ray tube and detector units, moves horizontally (in the x-direction) with a range of 1.9 cm in about 6000 motor steps. Two SLO-SYN stepping motors (M061-FD08), which are connected via a geared pulley and toothed belt to a precision ball screw, drive the front plate. The ball screw has a pitch of 1.27 mm (0.05") per

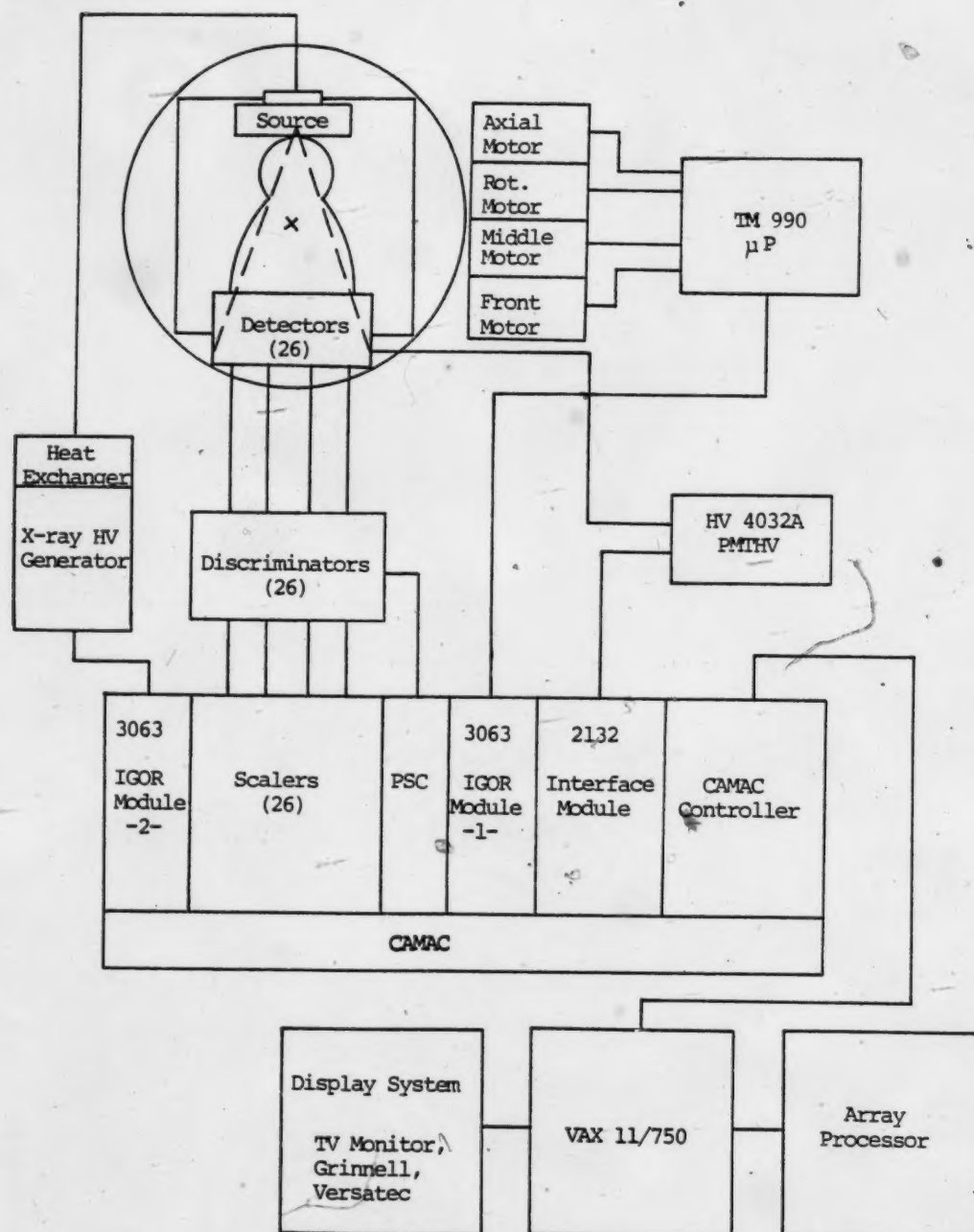


Figure 1. Components of the CT System.

revolution, and since the stepping motor (in half step mode) rotates once every 400 steps, the resolution of the front plate movement is $3.18 \mu\text{m}/\text{step}$. Every position in the 1.9 cm x-travel can be established with an accuracy of $\pm 1.9 \mu\text{m}$. This value includes cumulative errors in the drive system and resolution of the stepping motor.

2.1.2 Middle Plate

Movement of the middle plate is needed to optimize the relative geometrical resolution or "magnification" of the scanner. In order to utilize the entire X-ray beam, and hence maximize the relative geometrical resolution, the source/center-of-rotation (SCR) distance is decreased for small objects and increased for large objects. This plate, which supports the front plate, moves vertically (in the y-direction) with a range of 16.0 cm in about 25500 motor steps. The mechanical drive arrangement is similar to that of the front plate, except the two SLO-SYN stepping motors (M063-FD09E) are more powerful to accomodate the increased weight. The pitch of the ball screw is 2.54 mm (0.1") per revolution giving a positional resolution of $6.35 \mu\text{m}/\text{step}$. The accuracy is $\pm 6.8 \mu\text{m}$ over a 16.0 cm y-translational distance.

2.1.3 Rotational Plate

Movement of the rotational plate is activated during the collection of data. This plate, which supports both the

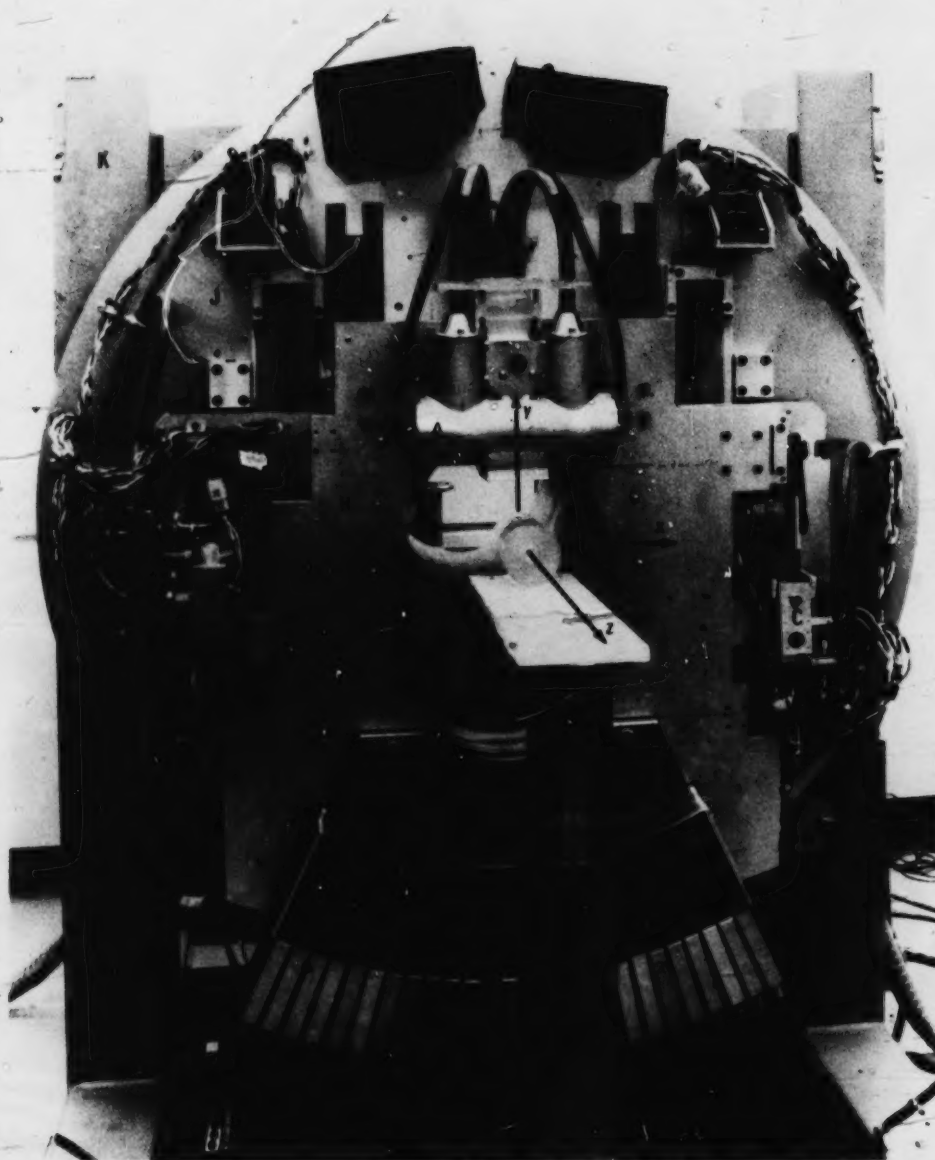


Figure 2. Photograph of CT Scanner in Reset Position.

(a) X-ray source, (b) 26-detector assembly, (c) air-brake, (d) rotational plate limit switch, (e) middle plate limit switch (front and axial limit switches not shown), (f) front plate motor, (g) middle plate motor (axial and rotational motors not shown), (h) front plate, (i) middle plate, (j) rotational plate, (k) stand.

front and middle plates, has a range of movement of 480° in about 24500 motor steps. A cross-section of the coupling arrangement between the SLO-SYN stepping motor (M112-FJ326) shaft and the rotational plate is shown in Figure 3. The resolution of the rotational plate is $(1/45.71) \times (360^\circ/400 \text{ steps}) = 0.0197^\circ/\text{step}$, and the accuracy (measured with a micrometer) is $\pm 0.001^\circ$ over a 360° travel distance.

A flexible coupling (Figure 4) was incorporated because it is not possible to exactly align the pulley and motor shafts. Misalignment of these two shafts causes a "wobbling" effect, which would induce excessive wear on the bearings of the rotational motor if the two shafts were fixed rigidly together. Furthermore, if the motor shaft was directly coupled to the small gear which interfaces with the rotational plate gear, the large inertia of the rotational plate, which must be overcome each time the motor starts or stops, would exert excessive torque on the gear. Accordingly, a special pulley belt device was designed [16] to couple the rotational plate with the motor shaft (Figure 5). Two spring/stop arrangements maintain tension in the pulley belt and ease the strain on the rotational motor shaft by partially absorbing the inertial forces accompanying any acceleration or deceleration.

2.1.4 Axial Movement

The third translational movement positions the entire scanner along the longitudinal axis (z-direction) of the

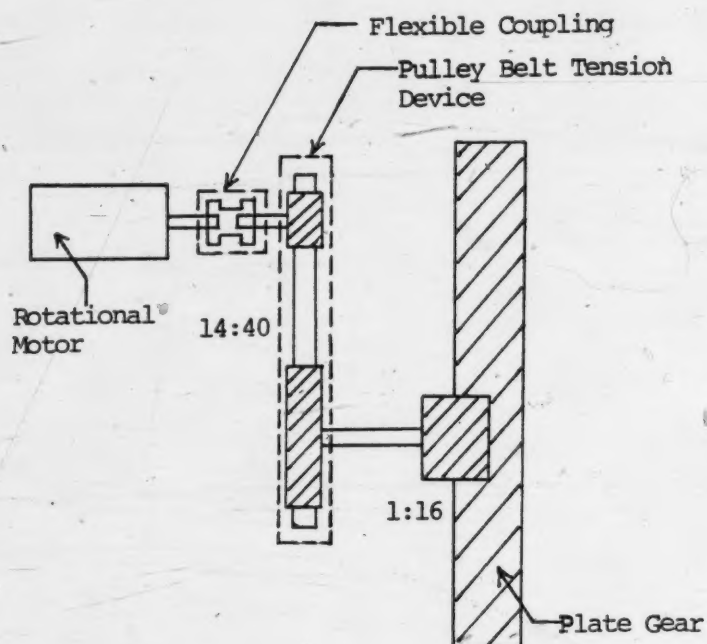


Figure 3. Cross-Section of Coupling Arrangement for the Rotational Plate.

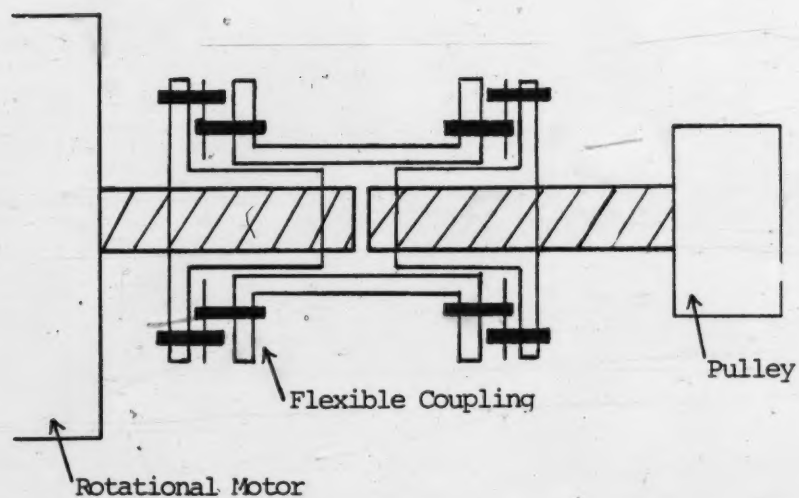


Figure 4. Flexible Coupling.

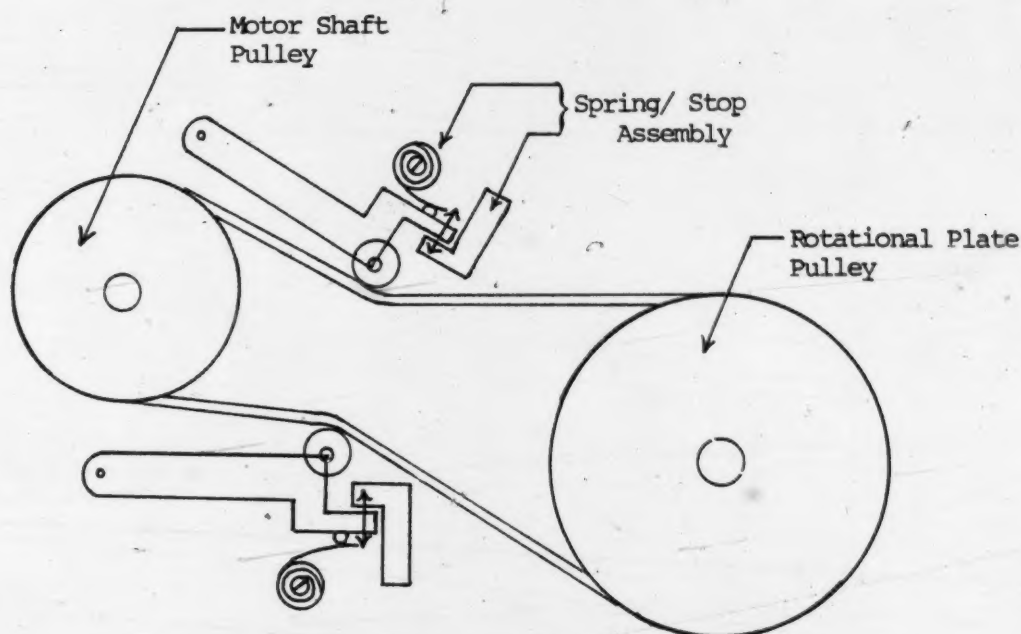


Figure 5. Pulley-Belt Device.

object and has a range of 75.0 cm in 47000 motor steps. The mechanical drive for this movement is one large SLO-SYN stepping motor (M112-FD12E) attached through a geared pulley and toothed belt to a precision ball screw with a pitch of 6.35 mm (0.25") per revolution, and a resolution of 15.83 $\mu\text{m}/\text{step}$. Every position in the 75.0 cm z-travel can be established with a precision of $\pm 15.5 \mu\text{m}$.

2.1.5 Torque/Speed Curves

The torque of a SLO-SYN stepping motor decreases as the stepping frequency increases (Figure 6a) [17,18]. This fact influences the performance curves of each of the four motors (Figure 6b) which are dependent upon the torque

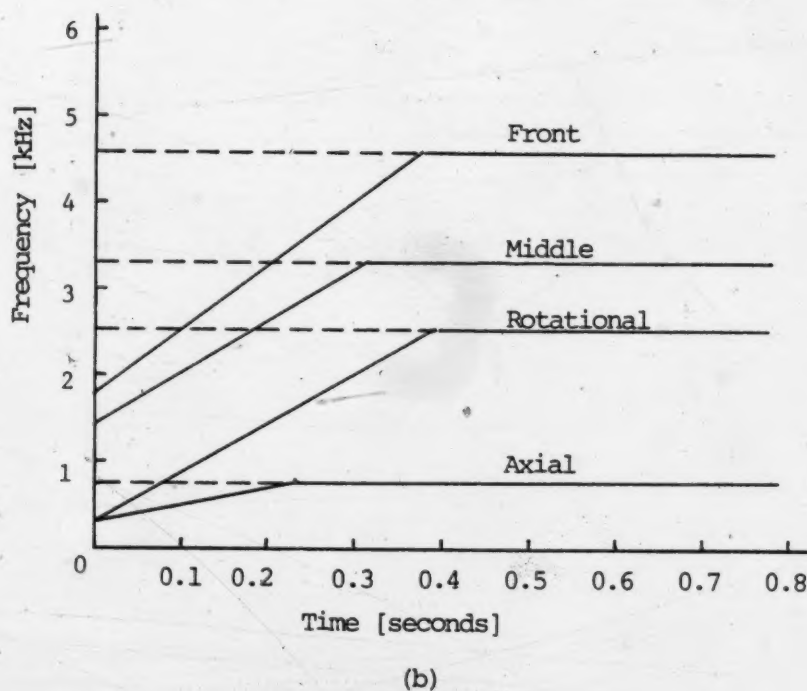
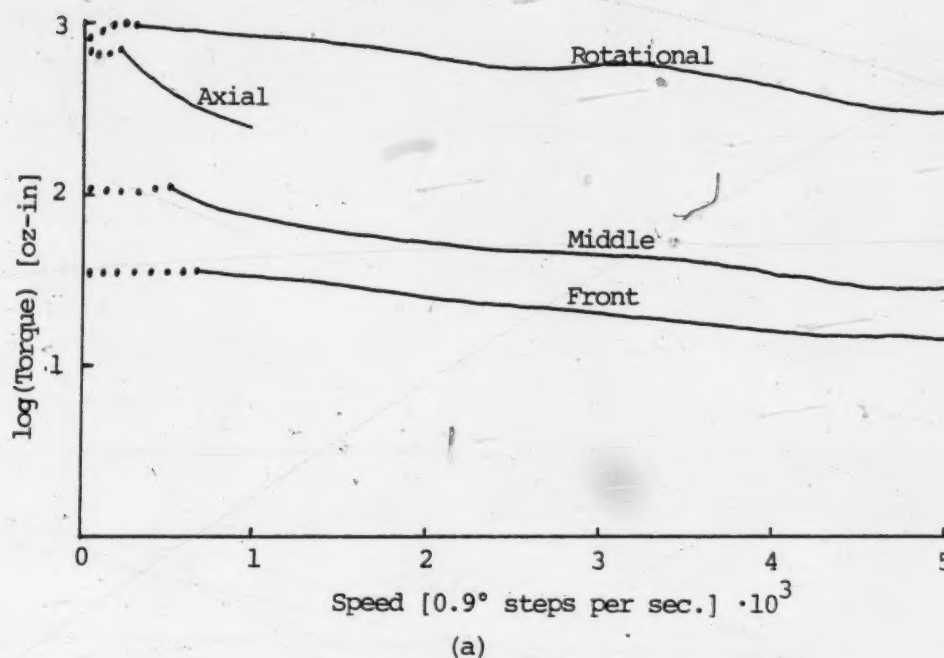


Figure 6. (a) Torque/Speed Curves, (b) Performance Curves.

The starting motor frequency and the maximum motor frequency depend on the torque specifications and load for that motor.

specification of the motor and on the load or total weight the motor moves. The maximum attainable frequencies for the front, middle, rotational, and axial motors are 4600 Hz, 3300 Hz, 2500 Hz, and 800 Hz, respectively. The maximum rotational frequency is the most important because it ultimately limits scan times. The part of the torque/speed curves represented with a dotted line is an area of possible motor resonance. In the case of the rotational motor, resonance occurs at about 175 Hz : Any frequency below 175 Hz causes the entire scanner to vibrate, which ultimately leads to poor image quality. Suitable acceleration/ deceleration ramps and motor speeds for a scan were determined based on these performance curves.

2.1.6 Position Switches

Position switches are required to determine a starting location for each of the scanner's movements. The principal element of such a switch is an oscillator which radiates a "sensing" field. When a metallic object, attached to a moving plate, enters the sensing field, eddy currents are induced in the object. This effect changes the internal impedance of the oscillator which provides a useable signal output from the oscillator circuit. In this manner, a geometrical zero point for each motor movement can be defined. The important feature of these position switches is their precision. It was experimentally determined that the position sensors switched reliably to increments of at least

3.175 μm , the resolution of the front motor.

2.1.7 Mechanical Safety Features

If a scanner movement exceeds its safe range of motion, a limit switch/relay circuit shuts off power to that motor. Positioning of each limit switch was done such that under the most extreme conditions (i.e. maximum plate momentum), the plate would stop (through spring-loaded pistons for the translational movements) before any damage occurred.

The limit switch device for the rotational motor had to be designed to allow for a rotational range of greater than 360° : Data are collected over 360° , but acceleration and deceleration ramps are also required. Two gear teeth are attached to the rotational plate and cause the limit switch device (Figure 7) to rotate incrementally, twice for every 360° rotation of the plate. By appropriately positioning the two teeth on the rotational plate, any range of movement between 360° and 540° is possible.

The middle plate of the scanner has a large range of movement, so if power to this plate failed, it would be possible for the plate to move down into a patient's limb. A pair of solenoid air brakes are utilized to lock the plate position whenever there is a power failure, or whenever the plate is not moving.

Additional safety features are implemented in the software control of the system and these will be discussed in chapter 3.

Microswitch and Cam Assembly:

Rotational
Plate:

Front
Gear:

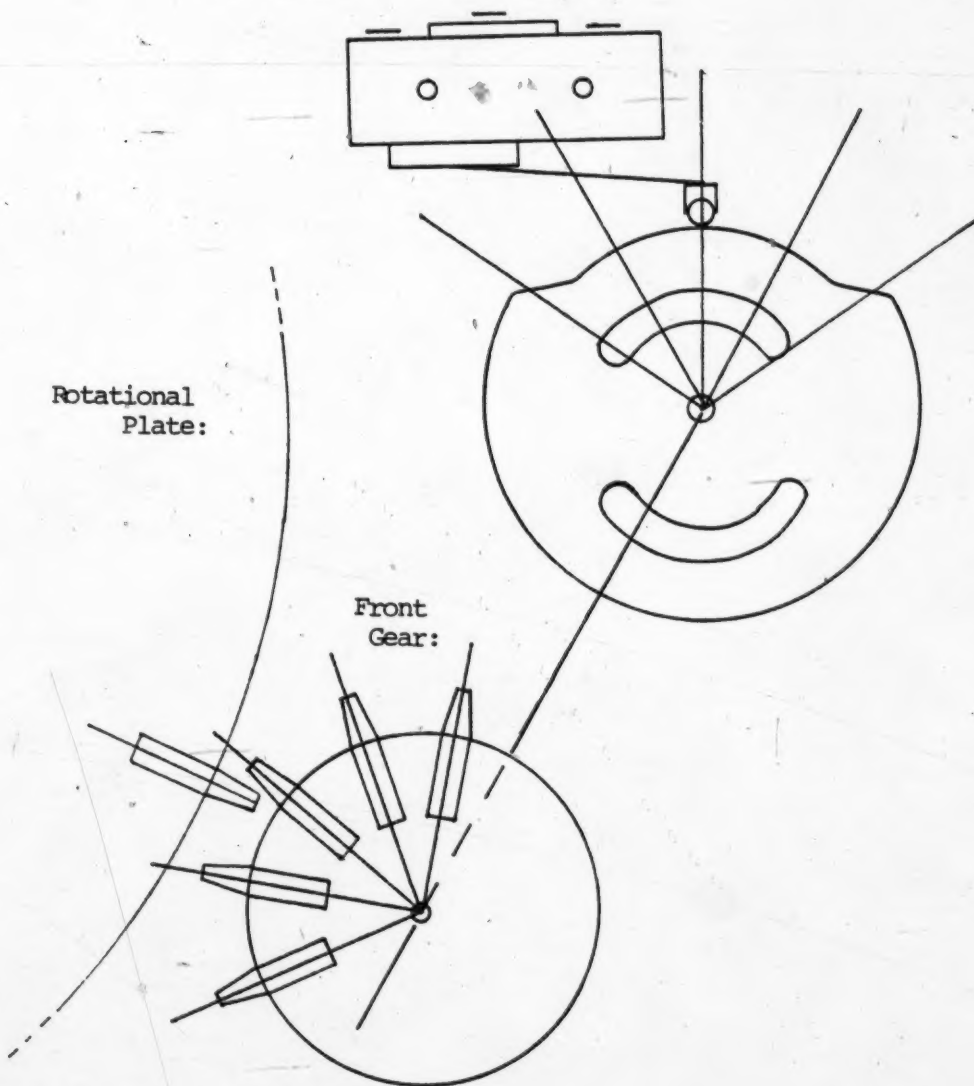


Figure 7. Rotational Limit Switch Device.

2.2 Control Rack

The control rack houses the power supplies, translator modules, relays, and control boards for each of the scanner's four movements, as well as the TM990/100MA microprocessor.

2.2.1 Translator Modules

Each SLO-SYN STM103 translator module is a printed circuit board containing the logic elements needed for bi-directional control of a stepping motor [17]. The axial, front, and middle motors of the scanner all use this type of translator module. The unit can be triggered by pulses from an internal oscillator or from an external pulse source, for example a microprocessor. Separate Kepco power supplies provide the regulated + 24 VDC and up to 10 amperes current necessary to operate the modules. The rotational motor, which must operate at higher frequencies and with a higher torque, utilizes a SLO-SYN TM600U combination power supply/translator module [18].

Each translator module can be operated in any of three modes: full-step with two windings on ($1.8^\circ/\text{step}$), full-step with one winding on ($1.8^\circ/\text{step}$), and half-step ($0.9^\circ/\text{step}$). All of the scanner motors operate in the half-step mode because of the finer positioning resolution, and because smoother operation at frequencies near the natural resonance points of the motor will reduce motion artifacts.

2.2.2 Translator Control Boards

The manual and remote control for each of the translator modules is handled via separate control boards. In manual mode, pulses are generated at specific frequencies by user control of the run and base speed potentiometers in addition to direction and speed select switches (Figure 8). Debouncer circuitry is used for generation of single motor steps. In remote mode, direction and speed control are handled via the microprocessor (μ P). The control board circuitry simply provides an interface between the μ P and the I/O of each translator module.

2.2.3 Relay Control Boards

As mentioned previously, if a scanner motor exceeds its range of safe movement, a limit switch opens and power to the corresponding relay is lost causing the motor to stop. In order to power the motor up again, a manual override switch is closed, which allows sufficient current to flow through a transistor to power the relay on (Figure 9). The scanner is then manually moved in such a way as to close the limit switch. As soon as this is done, the manual override switch is opened to allow the limit switch to function properly again.

The circuitry also shows how the air brake solenoid is linked to the power for the middle plate motor. When the limit switch for the middle motor is open, the MJ3001 transistor, which acts as a switching device, opens, and

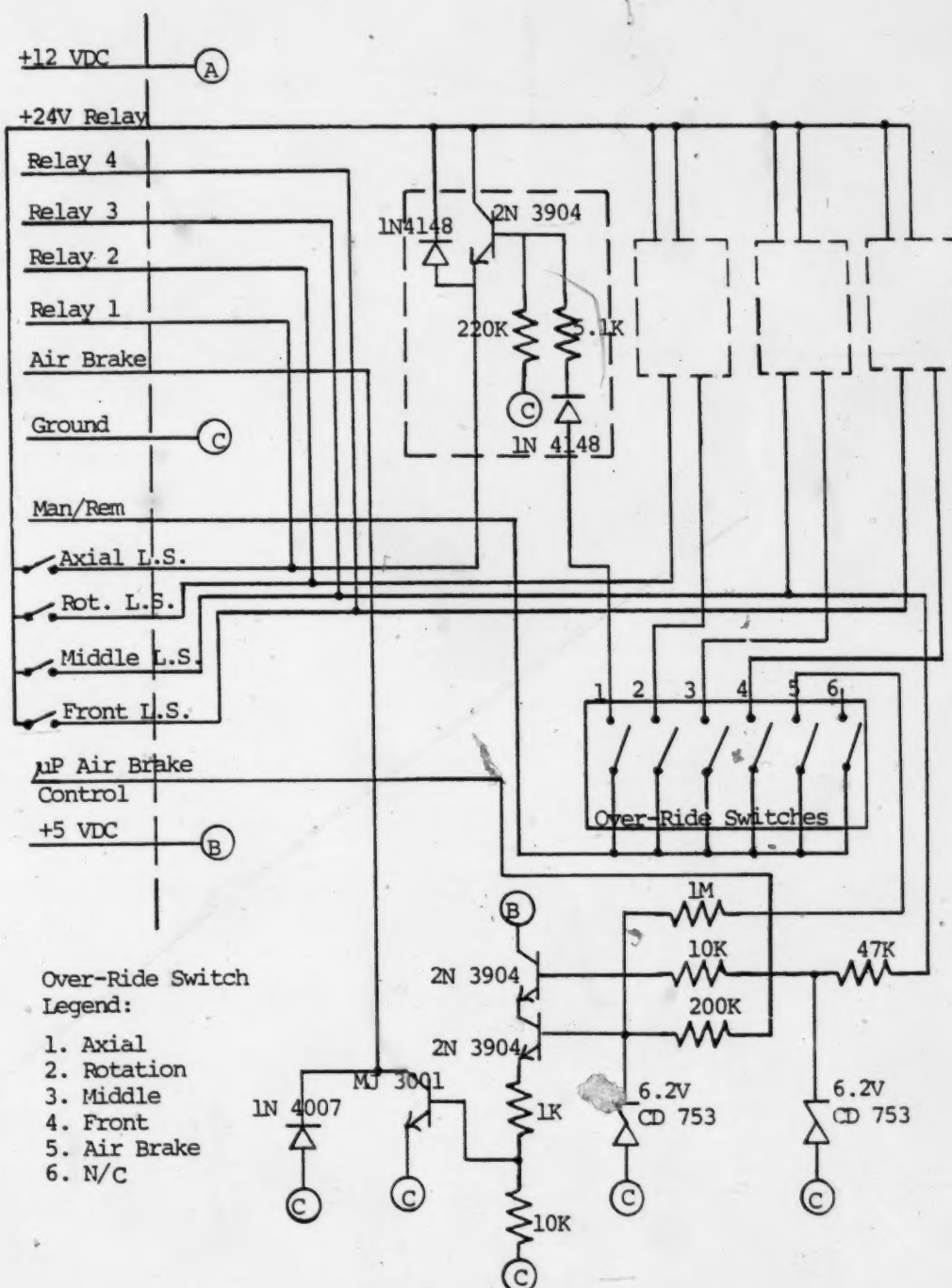


Figure 9. Relay Control Board Circuitry.

current will cease to flow through the air brake solenoid, causing the brake to lock the middle plate. The air brake can then be unlocked by activating the manual override switch for the middle plate. When operating the scanner via the μ P, the level of a control line from the μ P changes from low to high whenever the middle motor is used. This ensures that the air brake is unlocked whenever the middle plate is moved: As soon as the μ P has finished issuing steps to the middle motor, the air brake control line goes low and the air brake locks again.

Finally, if the limit switch for a plate opens under μ P control, power to that particular motor can only be re-activated under manual control.

2.2.4 Microprocessor

The Texas Instruments microcomputer system consists of a TM 9901/100 MA μ P board, an I/O board for communication with CAMAC, and a custom board containing special timers for running the 4 motors and status condition electronics. In addition, the microcomputer contains five TMS 9901 PSI I/O chips. Two are located on the μ P board and are associated with the position switches and with the timing and movement of the four scanner motors (see section 3.5). The other three TMS 9901 units are located on the I/O board; two are associated with data transfer between the VAX and the μ P, and the other with status conditions (see section 3.2).

The main functions of the microprocessor are the following:

1. To control the timing and movement of the four scanner motors.
2. To monitor the position of each of the scanner's motors through the use of proximity switches.
3. To monitor status levels for various devices including limit switches, air brakes, and power supplies for control circuitry and scanner motors. A series of LEDs attached to the front panel of the control rack are also used to monitor these status levels.
4. To control the timing of the interrupt pulses to the scaler units.

The microcomputer contains 4 Kbytes of RAM and 4 Kbytes of EPROM. For memory allocation in RAM and EPROM see Appendix III. The 4 Kbytes of RAM allow the user sufficient flexibility in choosing motor speeds and movements. Details of the μ P program are discussed in chapter 3.

2.3 Detection System

The scanner utilizes 26 detector units, each consisting of photomultiplier tube (PMT), scintillation crystal, and pre-amplifier circuitry. A Hamamatsu photomultiplier tube (R642-01) is used with a one-half inch diameter and one inch thick NaI(Tl) crystal. The PMT consists of 10 dynode stages with a maximum rating of 1250 VDC between anode and cathode.

The principal component of the pre-amplifier circuitry used in conjunction with the PMT is a LeCroy Model TRA1000 pre-amplifier (Figure 10a). The pre-amplifier acts as a current-to-voltage converter and features a closed loop gain of $20 \text{ mV}/\mu\text{A}$ [19] (for $R_f = 20 \text{ k}\Omega$), an inverting and non-inverting dual output, and excellent linearity over a range of output voltages from 0 to 1 V. Twisted pair wiring is used to carry the negative and positive going voltage pulses to the discriminators. A LeCroy 4415 discriminator unit sends a logic pulse to a LeCroy 4434 scaler unit if the pulse height is above a specified threshold level. An oscilloscope was used to determine the shape of a typical output pulse from the pre-amplifier circuitry (Figure 10b). The rise time (T_1) of 60 nsec is due primarily to the statistical spread of electron transit times in the PMT, and to the input impedance frequency response of the TRA 1000 [19]. The pulse length (T_2) of 640 nsec is due primarily to the decay time constant of the NaI(Tl) crystal (250 nsec).

The physical dimensions of the detector assembly are shown in Figure 11. The steel collimator is 5 cm deep with each collimator slot having dimensions of 1 x 3 mm. The collimator unit was precisely machined so that adjacent slots are spaced two degrees apart at a radius of 56 cm - the desired source/detector distance. To minimize background radiation reaching the NaI(Tl) crystals, the entire assembly surrounding the PMTs was shielded with 6 mm of lead. In

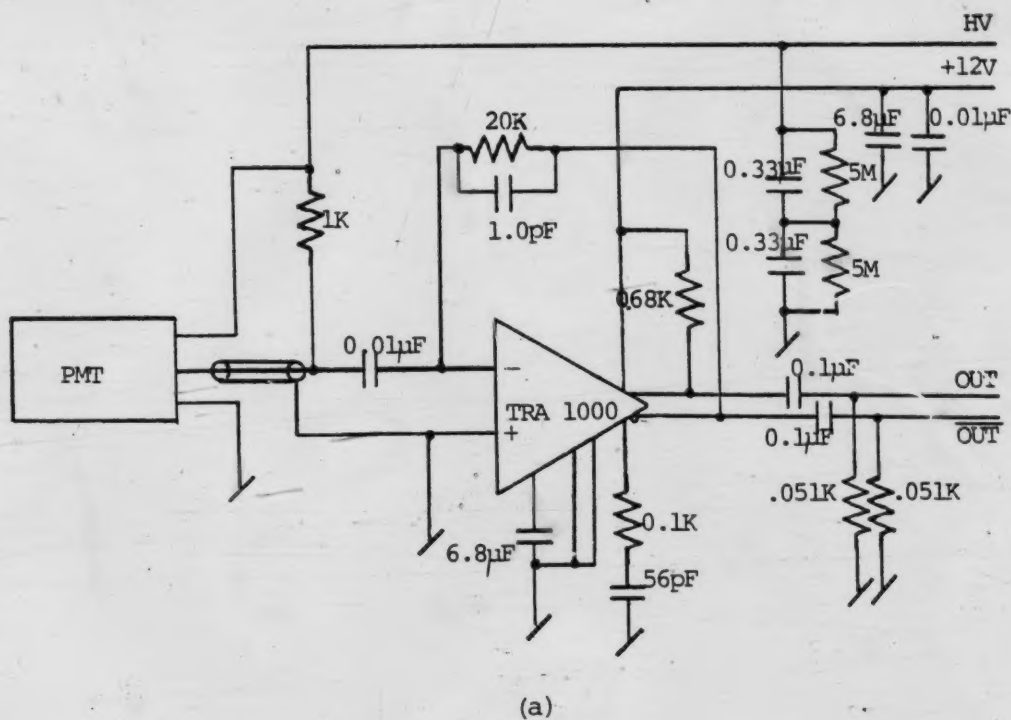


Figure 10. (a) Pre-amplifier Circuitry, (b) Typical Output Voltage Pulse.

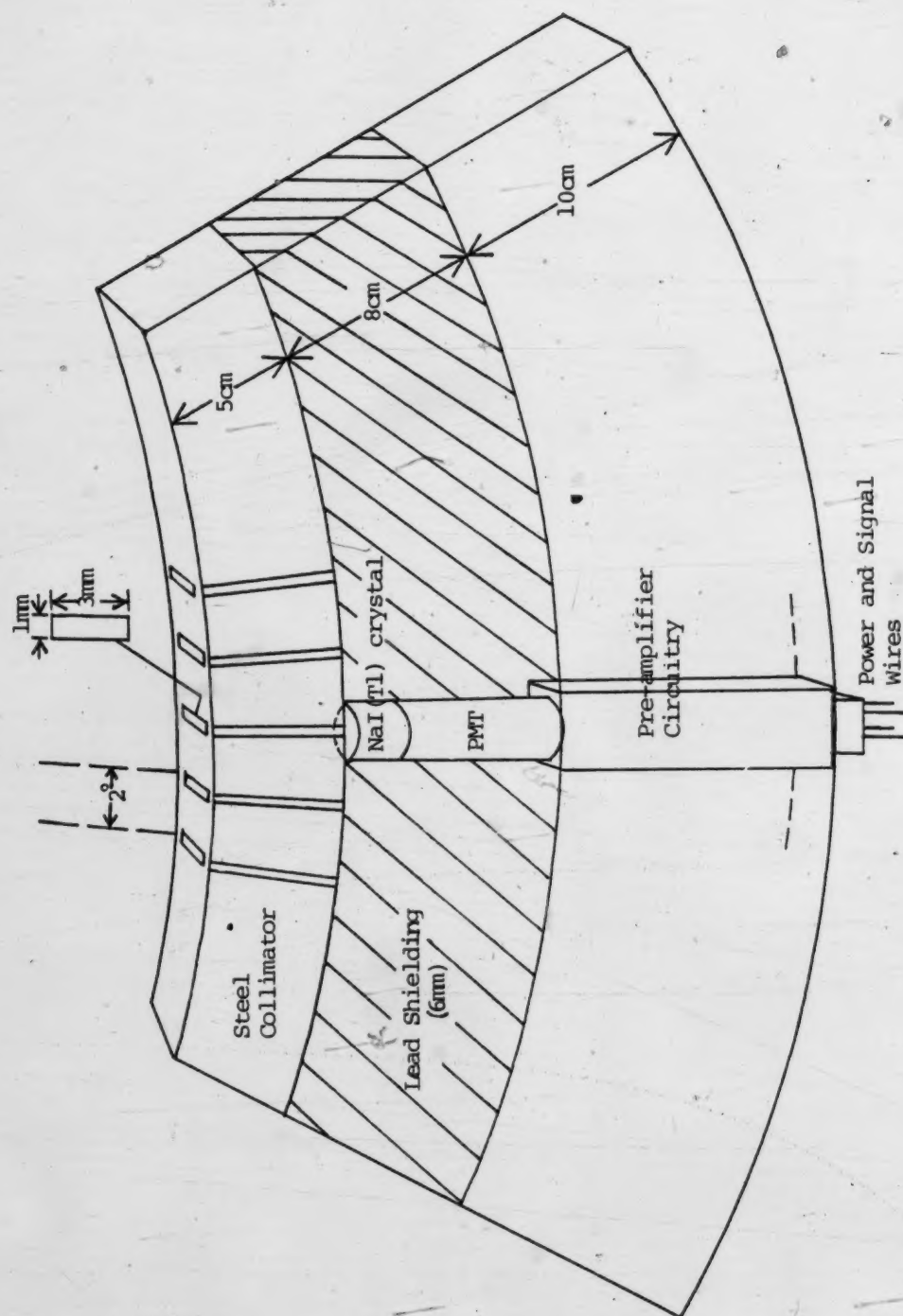


Figure 11. Detector Assembly.

addition, the detector assembly was electrically insulated from the rest of the scanner so that ground loops would not affect the output from the PMT/pre-amplifier circuitry. That is, motor pulses might produce unwanted pulses on the output line of the PMT if the PMT and scanner motor grounds were not separate.

2.4 X-ray Unit

A Comet X-ray tube (XL-85/100-1.0) was custom mounted with a useable X-ray beam angle of 50° . The cone of X-rays emitted from the anode is restricted to a fan beam by using a 3 mm thick lead collimator. A 0.70 mm copper pre-filter is used to "harden" the continuous spectrum of X-rays, thereby removing a significant portion of the unwanted low energy X-rays which would increase the dose to the patient without providing information. A diagram of X-ray intensity variation over the fan beam is shown in Figure 12. The X-ray flux decreases towards the outer detectors primarily because of the finite size of the fan beam. Flux is also reduced on the anode side of the fan beam due to the *heel* effect [20]; the beam near the heel, although of reduced intensity, is more penetrating because of self-filtration in the target. This X-ray tube has a maximum output of 100 kVp and 10 mA. The tube can operate continuously at 140 W. Power output above 140 W is limited to 16000 W-sec to avoid anode damage by overheating. The focal width of the anode is 1.28 mm. The X-ray tube is oil-cooled by a Comet heat exchanger (HX1000).

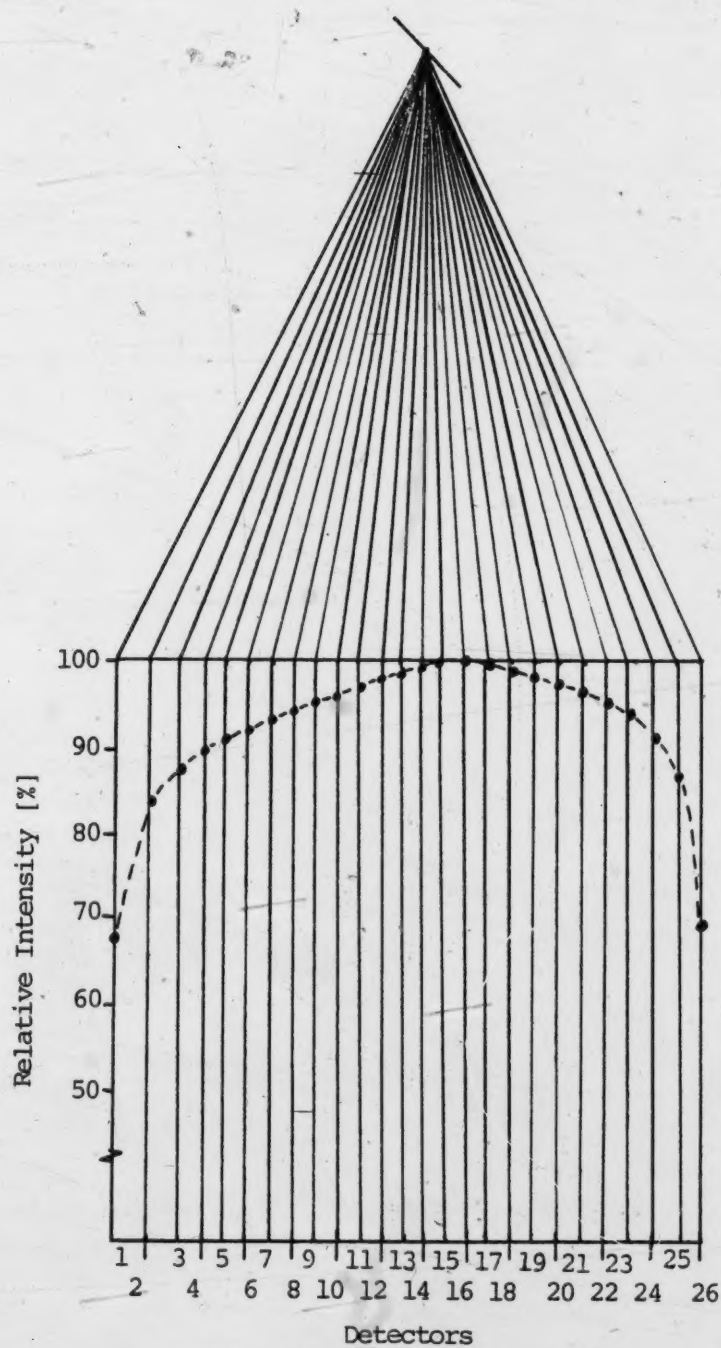


Figure 12. Intensity Spectrum.

Open count-rates were determined at each detector position in the fan beam and normalized to the detector position where the maximum count-rate occurs (detector 16).

A hermetically-sealed pump circulates the insulating oil from the X-ray unit to the radiator, where it is fan-cooled to ambient temperature. The cool oil then recirculates to the anode side of the X-ray unit. If the temperature of the oil exceeds 85°C a heat sensor switches the X-ray tube off.

A custom designed FUG X-ray generator (2HCN 600M-60000) is used with this system. The high voltage ripple is less than one part in 10^5 (peak-to-peak at 120 kV) or about 1.2 V, whereas the heating voltage ripple is less than 100 mV peak-to-peak [21]. By minimizing the ripple of the heating supply and high voltage supply, the associated error in the projection values is greatly reduced, thereby improving spatial and density resolution in the reconstructed image.

2.5 VAX 11/750 and CAMAC Interface System

In order to reconstruct images in a quick and efficient manner, a VAX 11/750 computer and CAMAC interface system are utilized. The VAX (Virtual Address Extension) system is a multi-user system which contains 3 Megabytes of main memory and 450 Megabytes of disc storage in addition to magnetic tape storage. An array processor (FPS 120-B) handles the various mathematical operations associated with the reconstruction process such as convolution and backprojection. A CAMAC (Computer Automated Measurement and Control) crate housing various modules provides the necessary interfacing between the VAX computer and the

TM990/100MA microprocessor, as well as the counting system and the X-ray generator. The CAMAC crate contains the following modules (Figure 1):

1. A Kinetic Systems crate controller module (3912) which communicates directly with both the VAX computer and the other CAMAC modules.
2. Two Kinetic Systems IGOR modules (3063) each featuring 16-bit input/output registers. One of the IGOR modules interfaces with the μ P and is used primarily to download parameter tables which control the scanner movements and to communicate any status messages which are sent by the μ P. The other IGOR module controls programming of the X-ray tube generator, allowing X-ray tube voltages and currents to be set through a VAX computer terminal.
3. A Kinetic Systems Power Supply Controller (PSC) module (3162-P1B) provides two analog output voltages via 12-bit digital-to-analog converters. For this system, the PSC is used to program the threshold voltage levels of the two discriminator modules.
4. Two LeCroy discriminator modules (4415) each consisting of a 16 channel, differential input discriminator. The common threshold level of the discriminator, with a range of -30 mV to -600 mV [22], can be set manually via front panel screwdriver control, or via an external voltage applied to the threshold control connector.
5. A LeCroy scaler module (4434) containing 32 channels of 24-bit scalars. Each scaler counts signals which have a

duration of greater than 10 nanoseconds and a maximum frequency of 20 MHz [23].

6. A LeCroy interface module (2132) allows bi-directional communication between the CAMAC Dataway and up to 15 PMT high voltage supplies (HV 4032A). The model HV4032A is a complete high voltage system providing up to 32 independently programmable channels. The PMT high voltages can be set manually or remotely via the 2132 module. The voltages for each channel are stored in the HV 4032A μ P and may be retained for up to 24 hours [24] without AC power, via a backup battery; hence, data will not be lost if power is temporarily turned off.

Other peripheral devices associated with the system are a Grinnell (GMR 270) imaging system, a black and white and a color TV monitor, and a Versatec (V80) hardcopy unit.

3. Software

All communication between the VAX and the μ P is handled through the CAMAC system: The VAX communicates with all CAMAC modules through the CAMAC controller, whereas the μ P communicates only with an IGOR module and with the scaler module. The procedure by which each word of data is transmitted from the I/O board of the μ P to the IGOR module of the CAMAC crate or vice-versa is the so called "handshake" method: The initiator of a message waits for a satisfactory/transfer message from the receiver before continuing. Four interrupt lines from the IGOR module to the I/O board are used for the handshake procedure. If the initiator is the VAX, the DOPR line is set and the DOAC line from the μ P responds; on the other hand, if the initiator is the μ P, the DIPR line is set and the DIAC line from the VAX responds. An unavoidable problem arises here in that the receiver, upon setting the data accepted line, assumes the initiator receives the message. If the initiator does not receive the message, the communication procedure breaks down with the receiver waiting for the next message from the initiator and the initiator waiting for the original response from the receiver. To prevent the system from "hanging up", a ten second time-out is implemented whenever communication is lost between the VAX and the μ P.

A list of μ P subroutines, as well as memory allocation in RAM is detailed in Appendix III.

3.1 Data Transfer Integrity

All data transfers between the VAX and the μ P are preceded by two hexadecimal words '0000' and 'FFFF' and followed by a CHECKSUM word, so, for instance, a 20-byte message actually requires 26 bytes to be transmitted. This is the second level of communication; the first level being the handshake procedure which occurs for each word in a message code. The format is as follows:

0000, FFFF, Transfer Data, CHECKSUM

where CHECKSUM is the sum of the transfer data values, and the words '0000' and 'FFFF' give an extra measure of data integrity by ensuring that all data lines can be driven low or high. For example, after a program section has been successfully executed, the μ P sends a continue message (3 0 0) to the VAX. The complete code of the message is

0000, FFFF, 0003, 0000, 0000, 0003

The VAX sums the three data values and compares the result to CHECKSUM. If the calculated CHECKSUM is correct and the data lines can be driven high or low, an O.K. message (0 0 0) is sent to the μ P. On the other hand, if an error occurs in the data transmission, an error message (F 0 0) is sent to the μ P. The μ P responds by re-transmitting the original message up to a maximum of three times. If the

error still persists, the μ P turns off any motors which were operating, as a result of lost communication.

3.2 Message Codes

All message codes are three words in length, but take six words for transmission because of the extra three words necessary to check data integrity. The various messages which can be sent to the VAX from the μ P are as follows:

1. F 0 0. Error message: results from incorrect transmission of data.
2. 0 0 0. O.K. message: results from a successful transfer of data.
3. 1 STATUS 1. Status message: indicates a mechanical/electrical fault in the system (Table 1).
4. 3 0 0. Continue message: occurs after a program section has been successfully completed and new input from the VAX is needed.
5. 4 MOTOR# POS. Position register message: occurs when the contents of a motor position register are uploaded to VAX (see section 3.9).
6. 5 MOTOR# 0. Early start motor message: results when an additional motor movement begins before another movement has completed. That is, it occurs when the STSNM register decrements to zero and the address mask for the new motor timer is removed (see section 3.5).

Table 1. Status Error Message Code

Bit#	Function
0	logic 1
1	logic 1
2	logic 1
3	logic 1
4	logic 1
5	Cooling Fans
6	logic 1
7	Power Axial Motor
8	Power Rotational Motor
9	Power Middle Motor
10	Power Front Motor
11	μ P Power
12	24VDC Relay Power
13	Rot. Motor Temp. Sensor
14	Remote/Manual
15	Brake

All bits in the status error code are normally set to logic 1, hence the normal status code is 'FFFF'. In the event of an error, the VAX program interprets the status error message code and displays the error on the user terminal. For example, if bit 9 goes low, the status error message 'FDFF' will be interpreted to state that there is a power failure to the middle motor.

The various messages which can be sent to the μ P from the VAX are as follows:

1. F 0 0. Error message.
2. 0 0 0. O.K. message.
3. 1 WP PC. Start program execution message: results in the downloading of a workspace pointer (WP) and program counter (PC) for a μ P program.
4. 2 LENGTH ADDRESS. Download message to μ P: specifies length and starting address of data or program placed in RAM.
5. 3 LENGTH ADDRESS. Upload message to μ P: specifies the

length and starting address in RAM for uploading data or program from the μ P.

3.3 Scan Initialization

Each time the μ P is reset or powered up, a hardware controlled initialization procedure is implemented which clears various μ P registers such as the position registers, and sets various interrupt lines to specific states. In order to perform a scan, the position of each of the four scanner motors must be determined relative to a corresponding position register (see section 3.9). Consequently, if the μ P has just been reset or powered up, a position register routine will have to be executed before a scan can be performed. The state of the position registers is tested with a VAX program through the use of a μ P power supply status flip-flop. The state of this status flip-flop is only correctly set after the position register routine is performed. Whenever power to the μ P is lost, the flip-flop status is changed so that the software control of the system does not allow data collection to proceed. In this manner, motor positions are known throughout all stages of the data collection procedure.

In addition to the μ P initialization procedure, several scan parameters such as number of detectors, number of scaler interrupts issued from the μ P, and number of rotations in a scan must be placed into a VAX file. Also, a common threshold level for all 26 discriminator units must

be entered via the PSC module, the PMT high voltages entered via the 2132-interface module and the HV4032A high voltage unit, and the X-ray generator kilovoltage and current entered via the second IGOR module. Once these initial steps are taken, a movement description table can be downloaded to the μ P and the collection procedure started.

3.4 Microprocessor Control of Scanner Movements

The data collection procedure begins by transferring a movement description table to a specified address in RAM. There are four possible transfer addresses, one for each scanner motor (see Appendix III). Each line of the movement description table contains three words of information. The first word contains the number of motor steps to be issued (STPCNT); the second word contains the motor pulse frequency (SPEED); and the third word contains the number of steps issued before an interrupt is sent to the scalers (SINT). The last line of any movement description table must have the value 0 0 0 in order to complete the program execution. In addition, each table contains acceleration and deceleration ramps where no data is collected, i.e. SINT=0.

Next, a WP and PC are downloaded to the μ P. The starting address for the workspace must be different for each scanner movement so that motors can be run simultaneously. The starting address for the PC indicates which motor is to be pulsed and in which direction (see Appendix III).

The STPCNT and SINT values from the first line of data are placed into the designated workspace, whereas the SPEED value is entered into the CRU unit of a TMS 9901 PSI chip to control the motor timing (see section 3.5). After each motor pulse is issued, STPCNT is decremented by one, and if SINT $\neq$ 0, it is also decremented by one. Each time SINT decrements to zero, an interrupt pulse is sent to the scalers, whereas each time STPCNT decrements to zero, a new line of data is placed into the μ P workspace and CRU unit. The procedure continues until the data line 0 0 0 is entered. At this point the μ P sends a 3 0 0 message to the VAX indicating a successful execution of the program. The μ P now returns to its wait state for further instructions from the VAX. It is important to note that during a motor movement, the READ lines of the VAX are enabled so that status messages from the μ P can be processed immediately. If a status error message is received during a motor movement, the VAX would generally send an interrupt pulse to the μ P telling it to stop all motors.

A common error occurs when the scan parameters entered to a file in the VAX and the movement description table downloaded to the μ P do not coincide. For example, if the VAX expects to receive 20 interrupt pulses from the scalers, whereas the μ P issues only 19 pulses, ~~then the VAX will time~~ out waiting for the 20th interrupt pulse from the μ P.

3.5 Motor Timing: Simultaneous Motor Movement

The SPEED value in Hertz is the frequency at which the μ P program will generate pulses to a particular scanner motor. However, the value actually sent to the TMS 9901 PSI unit is $23810/\text{SPEED}$. This value will now be explained: The μ P clock frequency is $\phi = 3.04768$ MHz. The TMS 9901 PSI unit when operated in clock mode decrements a given 14-bit number in its register at a rate of $\phi/64 = 47620$ Hz. When this number is decremented to zero an interrupt is generated. A 14-bit number has values ranging from 1 to 16384 hence the interval time between interrupts has a range from 21.0 μ sec to 344 msec. If only one scanner motor was run at any given time, this TMS 9901 PSI unit would be sufficient to control motor timing. That is, $47620/\text{SPEED}$ would be the value entered into the TMS 9901 register.

Simultaneous motor movement requires four special timer units (74LS374), an address decoder (74LS138), and a frequency divider (74LS393) in addition to the PSI unit (Figure 13). The additional timers act as frequency dividers so that each motor can run at a separate frequency.

The PSI unit is now used to enter 16-bit numbers into the registers of each of the four timer units. Values are selectively placed into individual timer units through the use of the address decoder. If a number is to be placed into a particular timer register, the corresponding load line from the address decoder unit is pulsed. The clock pulses to the timer units are sent via the 74LS393 chip which operates

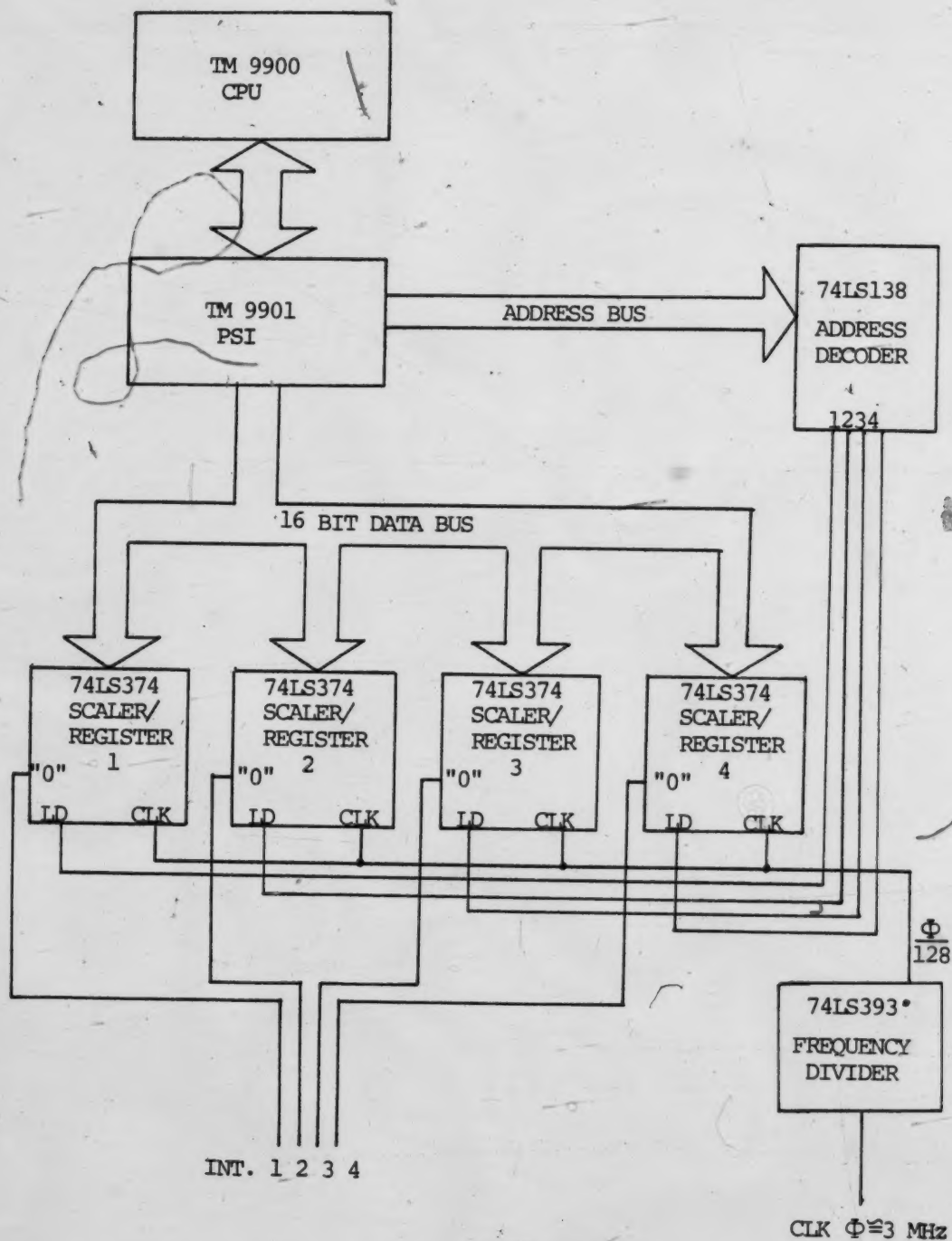


Figure 13. μ P Components Used to Control Motor Timing.

at a frequency of $\phi/128 = 23180$ Hz. Hence the values in the timer registers are decremented at a rate of 23810 Hz, and by loading different values into the timer registers, different pulse-rates will occur for each activated motor; a pulse is sent to a specific motor whenever that motor's timer register decrements to zero. In order to pulse a motor at a frequency SPEED, the scaler timer unit needs to be downloaded with the number $23810/\text{SPEED}$.

In summary, the 74LS393 chip pulses all timer units simultaneously at a frequency of 23810 Hz. A timer unit is activated by selectively downloading a particular number into its register through the use of the PSI and decoder units. That register number ($23810/\text{SPEED}$) dictates the pulse frequency for a particular motor. Motors can be run simultaneously by activating more than one timer unit. Additional motors can be started at specific times in the pulse sequence of another motor by using the STSNM register.

3.6 Data Processing: An Example

To illustrate some of these data processing principles, consider the movement description table SCAN.000 (Table 2). In order to activate the rotational motor, this table is specified to memory location starting at F25E. The download message to the μP is

0000, FFFF, 0002, 001E, F25E, F27E

Table 2. Movement Description File: SCAN.000

STPCNT	SPEED	SINT
5	200	0
10	300	0
15	400	0
50	500	5
15	400	0
10	300	0
5	200	0
0	0	0

STPCNT: Number of steps issued to motor

SPEED: Motor frequency

SINT: Number of steps before an interrupt is issued to the scalars (interval size).

where 001E is the length of the table in bytes. Upon receipt of the O.K. message (0 0 0) from the μ P, the VAX would download the data table starting at address F25E in RAM. After a successful transfer of the data table, the collection process can begin. WP and PC are sent to the μ P using the execute message code

0000, FFFF, 0001, FEE0, 0744, 0625

A PC equal to 0744 indicates that the rotational motor is to be pulsed in a CCW direction. The first line of data is accessed: STPCNT=5 and SINT=0 are placed in the specified workspace, whereas $23810/\text{SPEED} = 119$ is placed in the timer register for the rotational motor. STPCNT will be decremented to zero at a speed of 200 Hz, whereby the next data line is processed. The first three lines of data

accelerate the rotational motor up to the data collection frequency of 500 Hz. For the data line 50 500 5, a total of 50 pulses are sent to the rotational motor at a frequency of 500 Hz. After each pulse, both STPCNT and SINT are decremented by one. When SINT=0, a pulse is sent to the scalars, and SINT is reset to its initial value, 5. Hence 10 pulses are sent to the scalars at a frequency of 100 per second. By varying the SINT parameter, the interval or distance between projection points can be changed. Subsequent data lines are processed in the above manner until the data line 0 0 0 is reached. At this point a successful program execution message (3 0 0) is sent to the VAX, which responds by returning to its normal state. The μ P is now ready to accept new instructions from the VAX.

In an actual scan data is collected every $2/3^\circ$, so over a 360° rotation, 540 interrupts are sent to the scalars. The resolution of the rotational plate is $0.0197^\circ/\text{step}$ (section 2.1.3), so in order to rotate the scanner exactly 360° , 18285.71 steps are needed. Therefore the average interval length is $\text{SINT} = 18285.71/540 = 33.86$ steps. This fraction is dealt with by alternating in appropriate proportions SINT values of 34 and 33 steps (Table 3, ROTATION.000). The sum of the STPCNT values, excluding acceleration/deceleration ramps, is $18286 + 66$ steps: The necessity for an extra 66 steps or 2 intervals is explained in section 3.7.

Table 3. Movement Description File: ROTATION.000

STPCNT	SPEED	SINT
5	200	0
10	300	0
15	400	0
99	500	33
1292	500	34
198	500	33
1258	500	34
198	500	33
1292	500	34
198	500	33
1258	500	34
198	500	33
1292	500	34
198	500	33
1258	500	34
132	500	33
612	500	34
132	500	33
1258	500	34
198	500	33
1292	500	34
198	500	33
1258	500	34
198	500	33
1292	500	34
198	500	33
1258	500	34
198	500	33
1292	500	34
66	500	33
33	500	0
15	400	0
10	300	0
5	200	0
0	0	0

An interrupt is sent to the scalers on average every 33.86 motor steps ($2/3^\circ$) by alternating SINT values between 33 and 34 steps/interrupt

3.7 Scalers

Each of the 32 channels of the scaler module collects data continuously until an interrupt pulse is issued from the μ P. At that point, the values in each channel register are dumped into a temporary buffer in the scaler module and the registers are cleared to begin collecting data for the next interval. The data placed in the scaler buffers are read into a temporary I/O buffer of the VAX. Data in this temporary buffer are transferred to the specified reconstruction data file after all interrupts have been issued for one scanner rotation. The I/O buffer of the VAX is limited to 64 Kbytes, hence if 26 detectors are collecting 4 bytes of data, a maximum of 630 interrupts can be issued for any one rotation. The time interval before the first scaler interrupt is unknown, hence the scaler channels will contain meaningless data for this first interrupt. All subsequent interrupts are issued at constant time intervals so useful data is obtained. The meaningless data from the first interrupt is compensated for by collecting data over an extra scan interval (Figure 14).

3.8 Emergency Stop

The μ P program also features an emergency stop routine for the motors. It is invoked on command from the VAX (the user presses any key on the terminal while the scanner is running), or as a consequence of the μ P losing communication with the VAX.

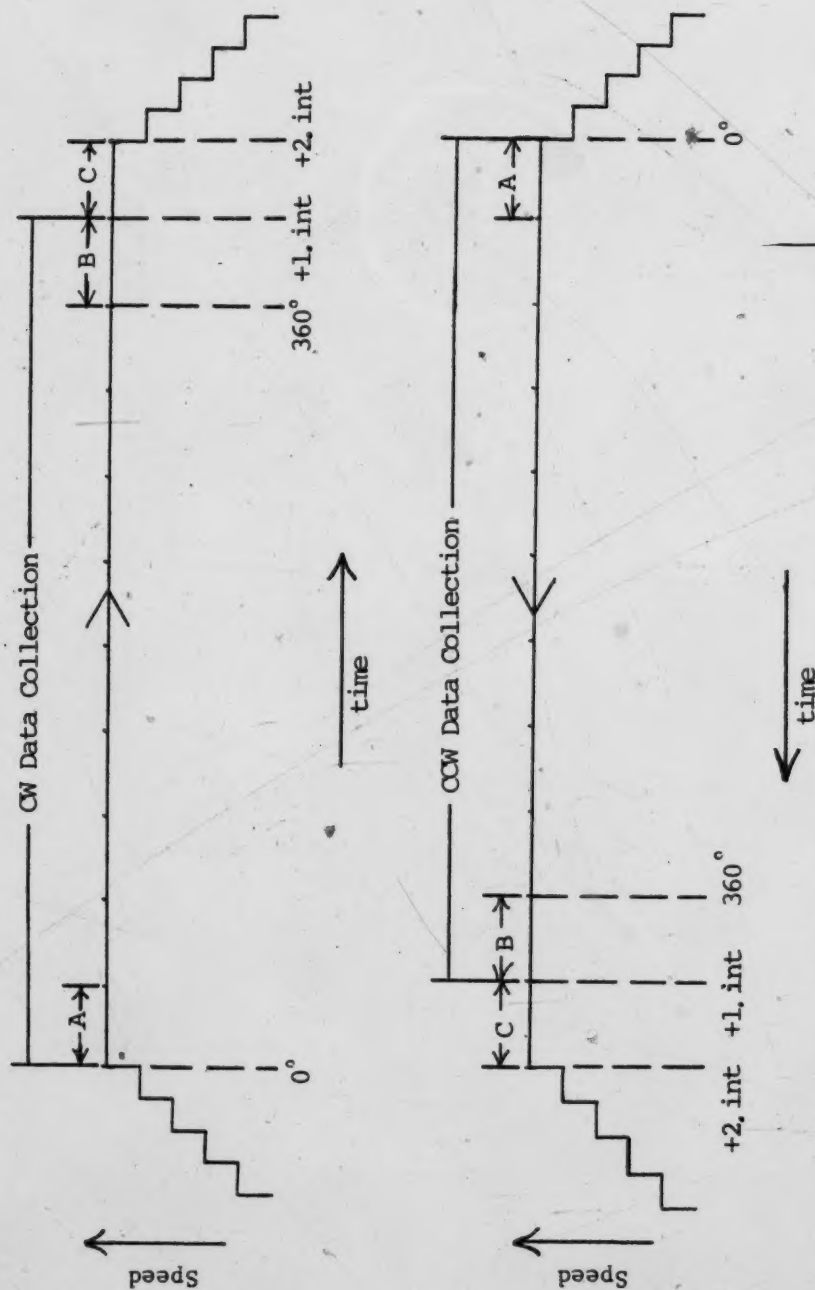


Figure 14. Rotational Movement for Data Collection.

The meaningless data collected over interval A is compensated for by collecting data over 360° plus one interval (B). The scanner rotates another interval (C) without collecting data before decelerating so that the 1st interval (A) of the reverse direction coincides with interval C of the first rotation. The step-like fashion of the acceleration/deceleration ramps is also shown.

When an emergency stop is issued, the program jumps to the next data line in the table which has a SPEED entry less than the present one. In other words, the start of the deceleration ramp of the motor table is placed in the workspace whenever an emergency stop occurs. Upon completion of the motor movement, the STSNM and SINT registers of the workspace are set to 'FFFF' so that no new motor may be started. In the event that more than one motor is running simultaneously, the emergency stopping sequence is performed for all motors in the order axial, rotational, middle, and front.

3.9 Motor Position Registers

Enabling of the position register for a given motor and then passing the motor through the corresponding proximity switch zeroes the position register at always the same plate location. This is an important feature of the system when the starting location for a multitude of scans is to be constant.

To show how the position of the axial motor can be determined relative to the corresponding proximity switch, consider the movement description table PROXIMITY.000 (Table 4). The axial motor moves according to PROXIMITY.000 until the motor sensor passes through the proximity switch. At that point the following position register message is sent to the VAX:

Table 4. Movement Description File: PROXIMITY.000

STPCNT	SPEED	SINT
5	200	0
10	300	0
15	400	0
20000	500	0
15	400	0
10	300	0
5	200	0
0	0	0

0000, FFFF, 0004, 002C, XXXX, XXXX#0030

where '002C' is the number for the axial motor position switch (see Appendix III), and 'XXXX' means the contents of the position register could be anything because the proximity switch register has only just been enabled.

The contents of the position register are now cleared so that in the future, the register contents will equal '0000' when the motor passes through the proximity switch. Next an emergency stop is called to halt the motor movement. In this example, the motor will stop 30+1 steps past the proximity switch. The 30 steps being a result of the deceleration ramp, and an extra step at 500 Hertz is taken before the emergency stop routine is implemented.

The user can determine the position of the motor by uploading the contents of the position register. The following upload message is issued by the VAX:

0000, FFFF, 0003, 0002, FF52, FF57

The μ P acknowledges the message and sends 2 bytes of data starting at the upload address for the axial motor (FF52), to a specified file in the VAX. From this file the decimal value of -31 can be read. In this example it was assumed that the motor passed through the proximity switch sensor at a frequency of 500 Hertz, otherwise the number of steps in the deceleration ramp would be different.

4. Calibration Procedures

The calibration of 26 detector units is a complicated and time-consuming process. The counting response of each detector unit must be corrected for dead-time, background counts, and beam-hardening effects. In addition, the effective energy of each X-ray spectrum must be matched, and a suitable discriminator threshold determined. Geometric factors such as SCR distance and detector angles are important parameters to the reconstruction process and must be determined with high accuracy. Finally, long and short-term stability of the system must be established so that image quality can be optimized for various "heat-up" periods of the X-ray tube.

4.1 Effective Energy of the X-ray Spectra

The discriminator threshold setting for this system is common to all 26 counting units, hence it is necessary to match the effective energy for each detector output spectrum in order to record approximately equal count-rates for equal incident energy flux. For this calibration procedure, the X-ray tube is operated at 50.0 kVp, and a 0.46 mm tin filter ~~is used~~ (instead of the 0.70 mm copper filter used for scans) so that the energy spectrum peaks can be measured in the discriminator voltage range of -30 mV to -600 mV. If the copper filter was used with the source operated at 50.0 kVp, the spectral peaks would occur outside of the discriminator range. (In actuality, spectral peaks using the

copper filter can be matched in the -30 mV to -600 mV range by reducing the PMT high voltages; however, this results in poor signal/noise discrimination - see section 4.2). It is assumed that the spectra matched with the tin filter will still be matched when the copper filter is used for actual scanning procedures.

The method involves measuring integral count-rates as a function of discriminator threshold. By subtracting the integral values from adjacent threshold levels, a differential spectrum is determined (Figure 15). By adjustment of PMT high voltages the spectral peaks of all detectors are matched. Due to the high cost of the detectors (approximately \$1000 each) they were purchased in two allotments. The "new" version of the detector has a better energy resolution than the "old" version. In addition, the new detectors have improved sensitivity which means these tubes can be operated at lower potentials, thus improving signal-to-noise ratios. The different spectrum shapes from the two types of detectors makes the calibration procedure that much more critical for reducing reconstruction errors.

An automated procedure to calibrate detector output spectra is implemented using the PSC and 2132-interface modules of the CAMAC unit (Figure 16). The peak position for each detector spectrum is compared to a calibration voltage level of -400 mV, and if the offset of the peak is greater than ± 10 mV, the PMT high voltage is adjusted:

$PMTHV(I) = PMTHV(I) - (Cal\ Factor(I) \cdot Peak\ Offset(I))$, where

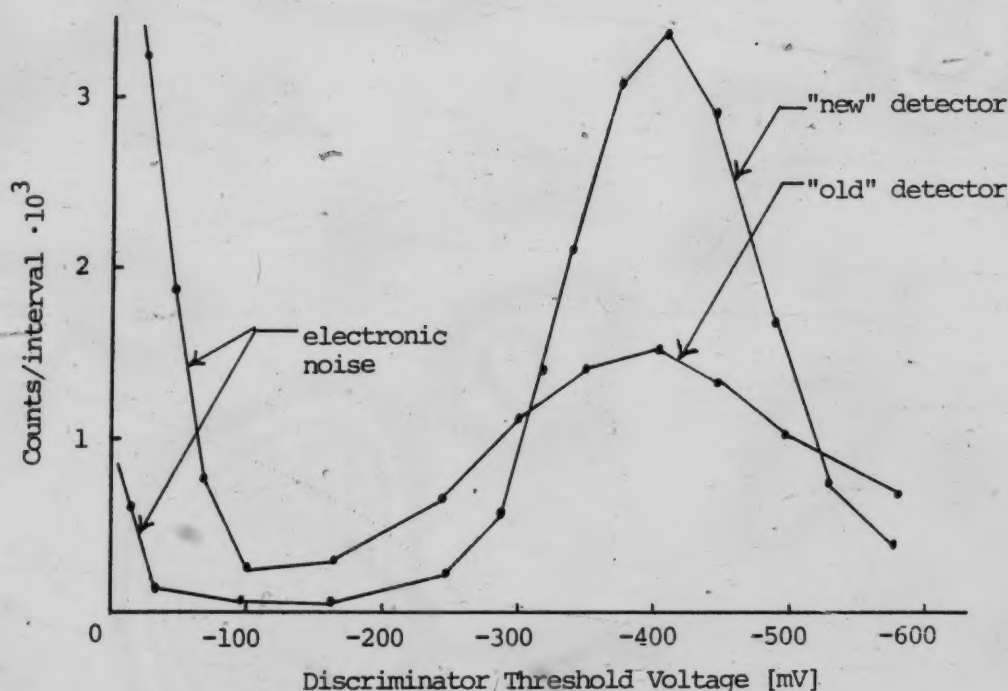


Figure 15. Differential X-ray Spectrum.

X-ray tube operated at 50.0 kVp with a 0.46 mm tin filter. The center of the spectrum peak is matched to a discriminator voltage level of -400 mV. Spectra are produced for the "new" and "old" version of detector. The new detectors have better energy resolution than the old detectors, and their improved sensitivity results in higher signal-to-noise ratios.

Cal Factor(I) relates a change in PMT high voltage to a change in spectrum peak position for the i-th detector. Cal Factors are listed in Appendix II. If a change in PMT high voltage for any of the detector units is necessary, the procedure to produce the voltage spectrums is repeated until all spectra have peak position values of -400 ± 10 mV. This procedure takes approximately five minutes per iteration and usually three iterations are needed. The stability of the system warrants running this calibration procedure once each

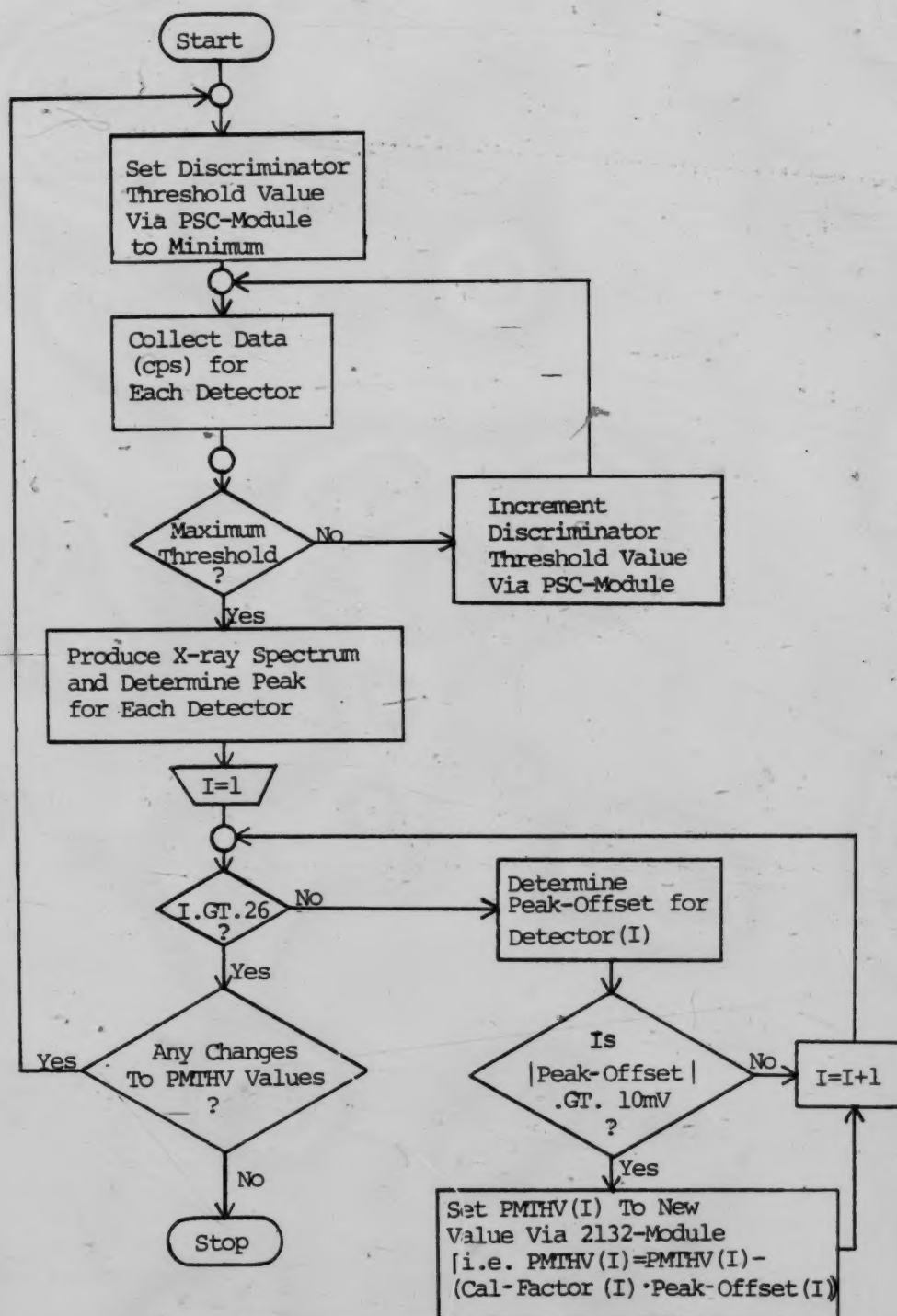


Figure 16. Flow Chart: X-ray Spectrum Calibration.

week (see section 6.4) however, the procedure is easy enough to be implemented at the start of each day.

4.2 Discriminator Threshold Level

A discriminator level needs to be established that can adequately remove most of the background noise without removing the signal. With all 26 X-ray spectra matched for peak position, a common discriminator threshold level can now be chosen to eliminate any low energy noise arising from detector units. The spectrum shape from the two detector versions differs significantly which adds to the problem of achieving a common discrimination level. Figure 17 shows how the differential spectrum shape changes as PMT high voltage is reduced for the two detector versions. At higher PMT voltages, the noise and signal energy separation becomes larger and so the discriminator level is easier to set. In addition, small changes in spectral response will have little or no effect on the counts recorded if the signal and noise components of the spectrum are widely separated. With this in mind, the PMT high voltage ranges were originally chosen, and after spectrum matching, a suitable discriminator threshold level was set at -200 mV; any pulse height smaller than -200 mV is rejected by the discriminator. The discriminator threshold level needs to be established at the start of the calibration procedure because dead-time, background, and beam-hardening corrections all are influenced by this setting.

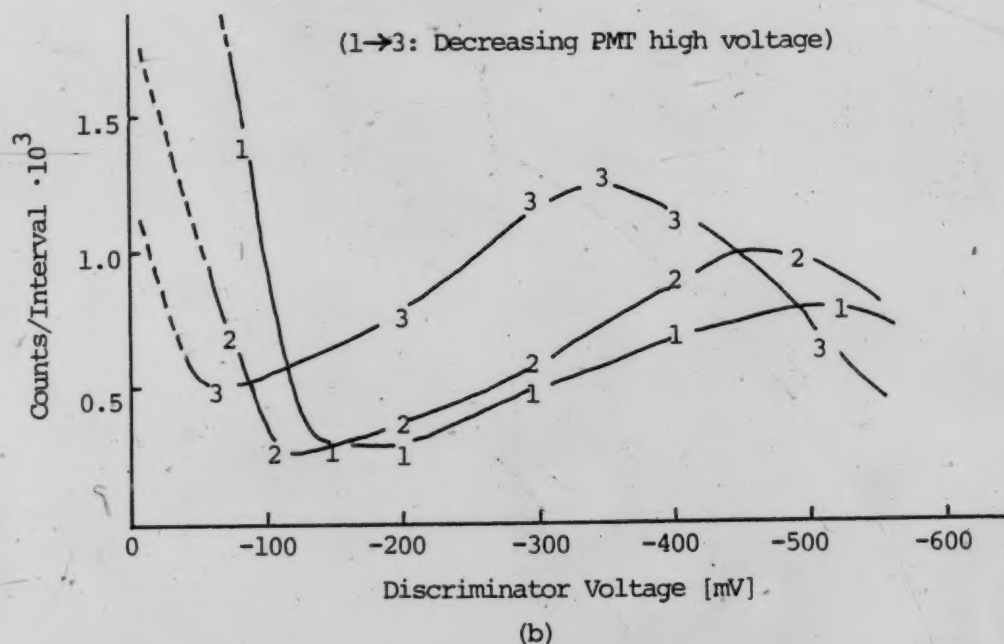
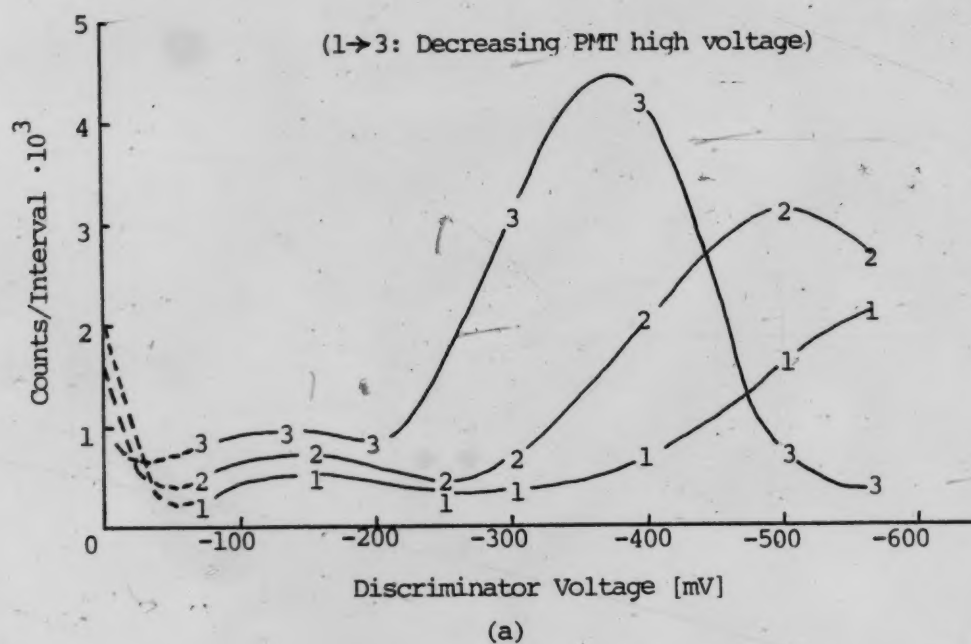


Figure 17. Differential Spectrum Response for Changing PMT High Voltage.

(a) New detector version, (b) old detector version.

4.3 Background Noise

Background noise consists of two components: a "dark-count" due to electronic noise, and a background radiation count. Figure 18 shows the combined background noise (counts/interval) as a function of discriminator voltage under various conditions. The response curves with (a) the tube off and shielded with 3 cm of lead and (b) with the tube on (50.0 kVp, 2.8 mA) and shielded are identical (for the old detector version) which indicates that the X-ray tube generator is not contributing to the noise spectrum. For the new detector version, no counts were recorded under these conditions.

Curves c and d show the noise response with the tube on (50.0 kVp, 2.8 mA) and no lead shielding (only the 0.70 mm copper filter) for the old and new detector versions, respectively. The number of counts recorded is two orders of magnitude larger than when the tube was shielded with lead (i.e. not recording a high photon flux). This increase is caused by a number of factors: (1) X-ray photons from the source are scattered in the copper filter by Compton interaction, and hence add to the counting spectrum as low-energy noise; (2) secondary interactions within the crystal could cause escape photons to arise, thus the recorded pulse has a reduced energy (however, low energy pulses recorded by this effect should be classified as "signal"); and (3) increases in two types of electronic noise [25]: (a) thermionic emission of electrons from the

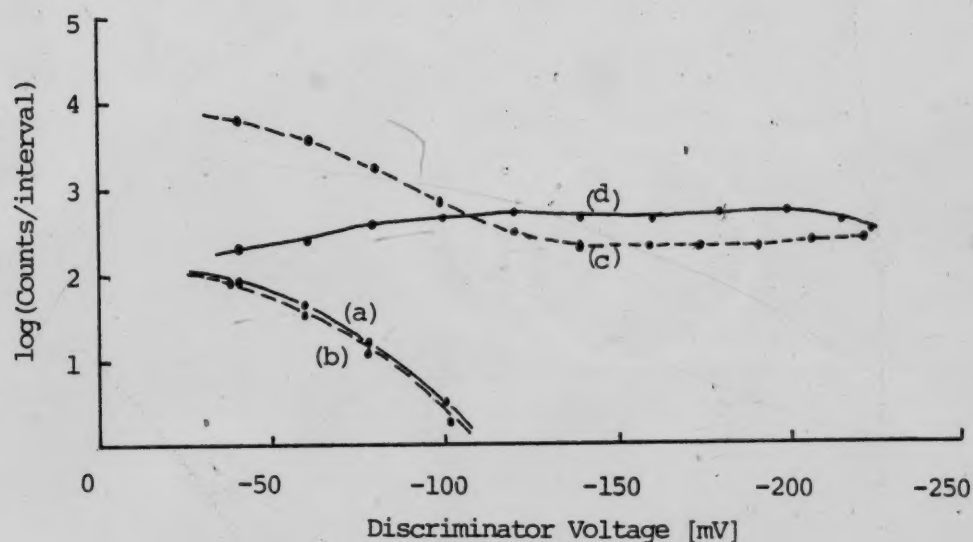


Figure 18. Background Noise Under Various Conditions.

Typical "old" detector: (a) tube off, 3 cm Pb shielding, 0.70 mm Cu-filter; (b) tube on (50 kVp, 2.8 mA) Pb and Cu-shielding; and (c) tube on, no Pb shielding. Typical "new" detector: (d) tube on (50 kVp, 2.8 mA) Cu-filter, no Pb shielding (n.b. negligible counts were recorded when tube was shielded with Pb). At discriminator voltage levels above -150 mV, a large contribution to the spectrum intensity is due to "signal".

photocathode or dynodes, and (b) ionization of residual gases inside the tube by electrons.

The majority of the low energy noise pulses can be removed by suitable discrimination at the output. However, higher energy noise passes through the discriminator units and adds to the noise of the signal. A background correction for these higher energy pulses, if existent, would involve a

correction proportional to the count-rate. The measurement of these pulses consists of recording count-rates before and immediately after a 3 cm lead attenuator is placed over the source. In this manner, the "active" background count-rate, caused by electronic noise, can be determined as a percentage of the open count-rate (n.b. this method does not account for noise caused by Compton scattered photons). At a threshold level of -200 mV, the average active background count-rate percentage is 0.0005% for the new detector version and 0.0015% for the old detector version. These values are identical to the background radiation percentages with the X-ray tube off. Any electronic noise greater than -200 mV is therefore independent of tube activity and can simply be expressed as an absolute count-rate. The background noise correction procedure involves reducing all count-rates recorded during a scan by, on average, 1.09 cps for the new detector version and 1.28 cps for the old detector version. The background count-rate above -200 mV is slightly lower for the new detector version because of the lower PMT high voltages. Appendix II shows the background count-rates for all detectors.

4.4 Dead-Time Corrections

All practical radiation detectors exhibit a continuously decreasing sensitivity with increasing intensity of radiation. The response curve (observed count-rate (N_o) versus photon flux) is thus not linear, as

it would be for a perfect instrument. With radiation counters, the non-linearity is due to the finite resolving time or interval that must separate two events if they are to be separately registered. For a NaI(Tl)/PMT detector unit, this "dead-time" is due to the fluorescence decay time constant of the crystal (about 250 nsec), and to the resolving capabilities of the pre-amplifier/amplifier and discriminator/scaler modules. This loss of counts is generally expressed for two cases. The "type 1" situation relates N_0 to the true count-rate (N_t) by the following expression [26]:

$$N_0 = N_t / (1 + N_t \cdot \tau_d) \quad (1)$$

where τ_d is dead-time. The "type 2" situation is given by the following expression [27]:

$$N_0 = N_t \cdot \exp(-\tau_d \cdot N_t) \quad (2)$$

Equation 1 assumes a constant dead-time after each recorded event, and equation 2 assumes a constant dead-time after every event, whether recorded or not. Figure 19 illustrates the various kinds of instrument responses.

To determine how to approach this dead-time problem, a plot of N_0 versus tube current, N_x (which is proportional to N_t) is produced for each detector of the specialized CT system (Figure 20). The experimental curve of Figure 20

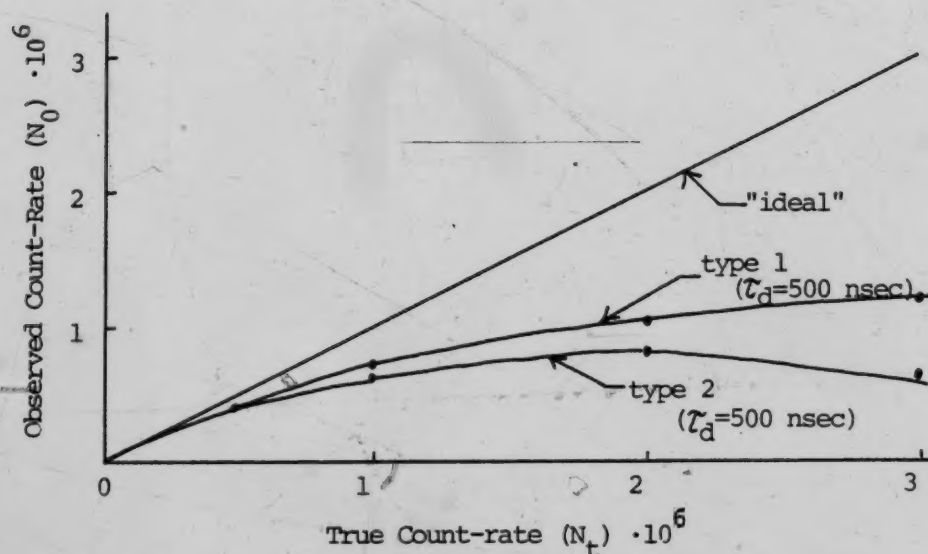


Figure 19. Counting Response Curves for Various Types of Systems.

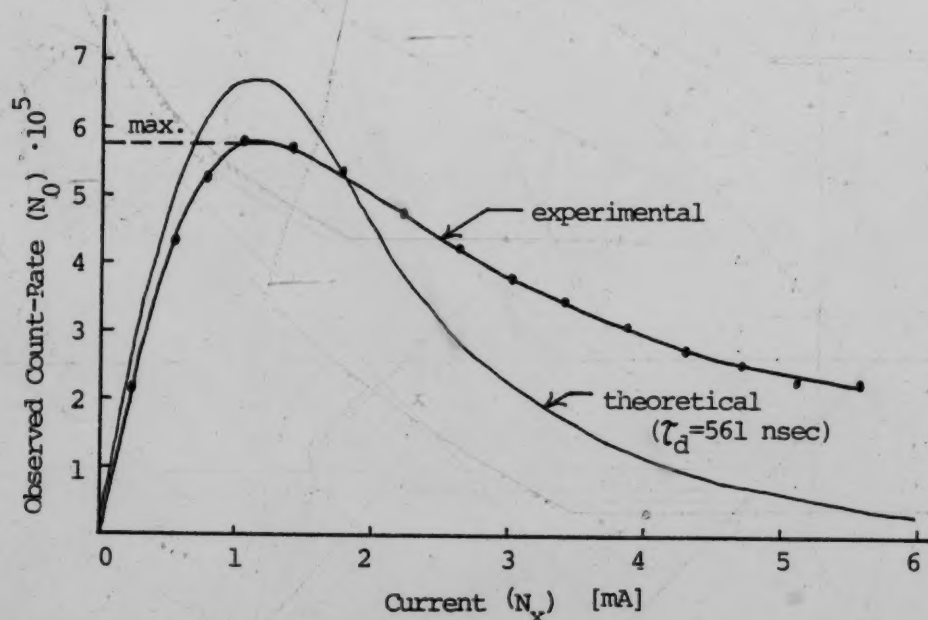


Figure 20. Counting Response Curve for a Typical Detector of the Specialized CT System.

A theoretical curve for a pure type 2 system ($\tau_d=561$ nsec) is used for a comparison, where τ_d was calculated from the maximum N_o value of the experimental curve.

suggests that the detection system has a type 2 response.

Differentiating equation 2 with respect to N_t gives

$$\begin{aligned} dN_o/dN_t|_{\max} &= -N_t\tau_d \exp(-N_t\tau_d) + \exp(-N_t\tau_d) = 0 \\ \Rightarrow N_t &= 1/\tau_d \\ \therefore \tau_d &= 1/(e \cdot N_o|_{\max}) \end{aligned} \quad (3)$$

Hence dead-time for a type 2 system can be calculated by determining the maximum observed count-rate for the system. Dead-time for the experimental curve is determined to be 561 nsec.

A theoretical curve ($\tau_d = 561$ nsec) was then plotted using a modified version of equation 2:

$$N_o = m \cdot N_x \exp(-\tau_d \cdot m \cdot N_x) \quad (4)$$

where m is a proportionality constant relating tube current to N_t . A value of m was estimated by assuming that $N_o \cong N_t = m \cdot N_x$ at low observed count-rates (i.e. where $\tau_d \cdot m \cdot N_x \ll 1$).

The different response of the theoretical curve indicates that this system is not a pure type 2 system; like most practical systems, it is a combination of type 1 and type 2 systems. The mathematical expression for this system is therefore complicated and solutions for determining the dead-time components of the system are difficult to obtain.

An easy but accurate method to correct for dead-time is to use a numerical approximation to the response curve [28].

Kohman has successfully used second, third, and higher order polynomial functions to correct for the non-linearity of various dead-time response curves. For this CT system, it can be shown (using an F-test [29]) that the experimental curve (in the region where the corrected count-rate response has a unique solution, i.e. below the maximum N_0 value) is sufficiently approximated by a second order polynomial (Figure 21):

$$N_0 = a \cdot N_x + b \cdot N_x^2 \quad (5)$$

$$\begin{aligned} \Rightarrow N_x &= \frac{-a \pm (a^2 + 4bN_0)^{1/2}}{2b} \\ &= -(a/2b) \pm ((a/2b)^2 + N_0/b)^{1/2} \end{aligned} \quad (6)$$

where a and b are constants to be determined. The relationship between the system response N_t and N_x for a perfect instrument is

$$N_t = m \cdot N_x \quad (7)$$

where m is the same proportionality constant used in equation 4. Substituting equation 6 into 7 gives:

$$N_t = m \cdot [-(a/2b) \pm ((a/2b)^2 + N_0/b)^{1/2}] \quad (8)$$

Taking first and second derivatives of equation 5 with respect to N_x , and observing the shape of the response curve (decreasing slope as tube current increases) indicates $b < 0$

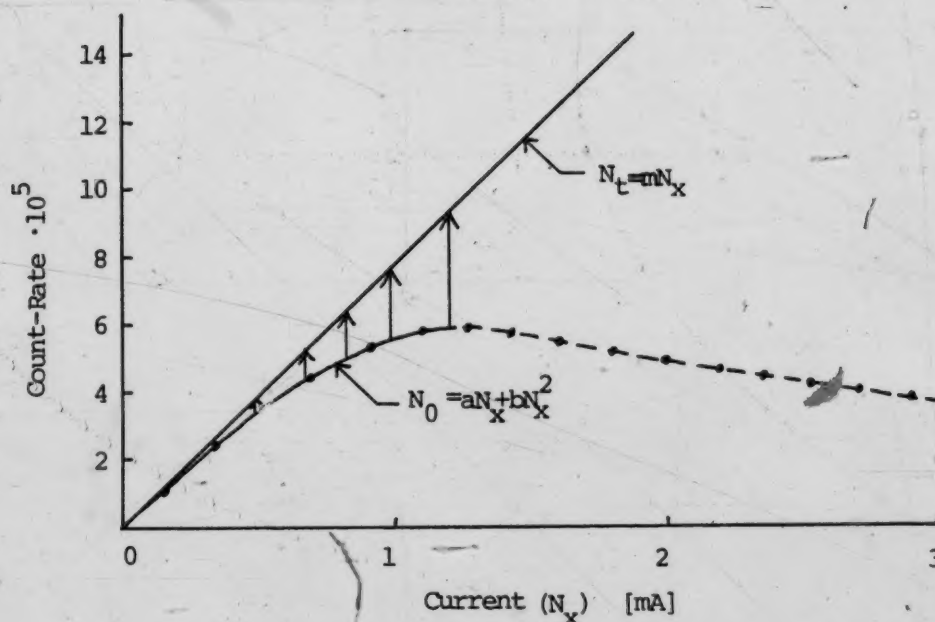


Figure 21. Numerical Correction of the Dead-Time Effect.

N_t is the count-rate for an ideal system, and N_x is any parameter proportional to true count-rate (e.g. tube current). The dashed line represents the region where a numerical solution to the experimental response curve is not unique.

and $a > 0$. Hence, at $N_0 = 0$, $N_t = 0$ only if

$$N_t = m \cdot [-(a/2b) - ((a/2b)^2 + N_0/b)^{1/2}] \quad (9)$$

Thus for any N_0 value, the corrected response N_t can be determined by solving for a and b (constants which are unique for each detector).

A least squares curve fitting procedure can be used to match the measured N_0 's and N_x 's of the experimental response curve to the curve $N_0 = aN_x + bN_x^2$, thereby determining a and b . The best line is based on the assumption that its location is such to make the sum of the least squares of the differences of the observed and the theoretical values a minimum. A weighted least-squares function for the second-order polynomial approximation is:

$$\sum_{i=1}^I \frac{(-N_{0i} + aN_{xi} + bN_{xi}^2)^2}{N_{0i}} = \text{minimum} \quad (10)$$

where I is the total number of experimental points (20 points were used to fit the curve). Taking partial derivatives of equation 10 with respect to a and b gives:

$$\begin{aligned} \frac{\partial}{\partial a} \sum_{i=1}^I \frac{(-N_{0i} + aN_{xi} + bN_{xi}^2)^2}{N_{0i}} &= 0 \\ \rightarrow \sum_{i=1}^I \frac{(bN_{xi}^2 + aN_{xi} - N_{xi}N_{0i})}{N_{0i}} &= 0 \end{aligned} \quad (11)$$

$$\begin{aligned} \frac{\partial}{\partial b} \sum_{i=1}^I \frac{(-N_{0i} + aN_{xi} + bN_{xi}^2)^2}{N_{0i}} &= 0 \\ \rightarrow \sum_{i=1}^I \frac{(bN_{xi}^2 + aN_{xi}^2 - N_{xi}^2N_{0i})}{N_{0i}} &= 0 \end{aligned} \quad (12)$$

The constants a and b can now be obtained:

$$a = \frac{\sum N_{x_i} \sum (N_{x_i}^2 / N_{o_i}) - \sum N_{x_i}^2 \sum (N_{x_i} / N_{o_i})}{\sum (N_{x_i}^2 / N_{o_i}) \sum (N_{x_i} / N_{o_i}) - (\sum N_{x_i} / N_{o_i})^2} \quad (13)$$

$$b = \frac{\sum N_{x_i}^2 \sum (N_{x_i} / N_{o_i}) - \sum N_{x_i} \sum (N_{x_i}^2 / N_{o_i})}{\sum (N_{x_i}^2 / N_{o_i}) \sum (N_{x_i} / N_{o_i}) - (\sum N_{x_i} / N_{o_i})^2} \quad (14)$$

$$\text{(where } \Sigma \equiv \sum_{i=1}^I \text{)}$$

By calculating a and b, equation 9 can be used to correct for the dead-time effect.

To test the method, the linearity of each detector response (after dead-time correction) is determined. On average, a linear correlation coefficient of $r=0.998$ ($p<0.0001$) was achieved. This method has the advantage of being easy and quick to implement. The raw data from the response curves (N_o versus tube current) can be obtained in less than half an hour. Dead-time corrections can therefore be easily updated to accomodate for electronic drift or anode wear.

4.5 Beam-Hardening Corrections

A beam-hardening correction is necessary whenever a polyenergetic photon beam is used. The reconstruction program assumes that the energy fluence spectrum remains unchanged as the beam passes through the imaged object; the effective energy of the beam remains constant. However, because the linear attenuation coefficient $\mu(E)$ depends on energy (e.g. $\mu \propto E^{-3}$ for a photoelectric interaction), the

energy fluence spectrum will change as it passes through matter. The higher energy photons whose linear attenuation coefficients are smaller will not be attenuated by the same extent as the lower energy photons, that is, the effective energy of the energy fluence spectrum will be increased (the beam will become "harder" or more penetrating) as the spectrum passes through the object. In other words, the effective linear attenuation coefficient, $\mu(E_{eff})$, will decrease in value as the polyenergetic photon beam passes through matter, when it should remain the same. An illustrative way of showing how beam hardening affects the reconstructed image is shown in Figure 22 [30]. If a profile of the CT numbers, $\mu(x,y)$, is made, homogeneous materials will have decreased CT numbers towards their centers. This "cupping" effect is primarily the result of beam-hardening (scattered or secondary radiation also causes cupping [31], but is low in a homogeneous object compared to the effect of beam-hardening) and degrades the precision and accuracy of the bone evaluation procedure.

An easy but imperfect means of correcting for this effect is to assume that the objects scanned will have a constant mass attenuation coefficient. Projection values were recorded as successive layers of "sandwiched" 1.65 mm thick aluminum and 6.0 mm thick plexiglass plates (materials whose mass attenuation coefficients closely match those of compact bone and soft tissue) were placed between the source and the detectors. From these data points, beam-hardening

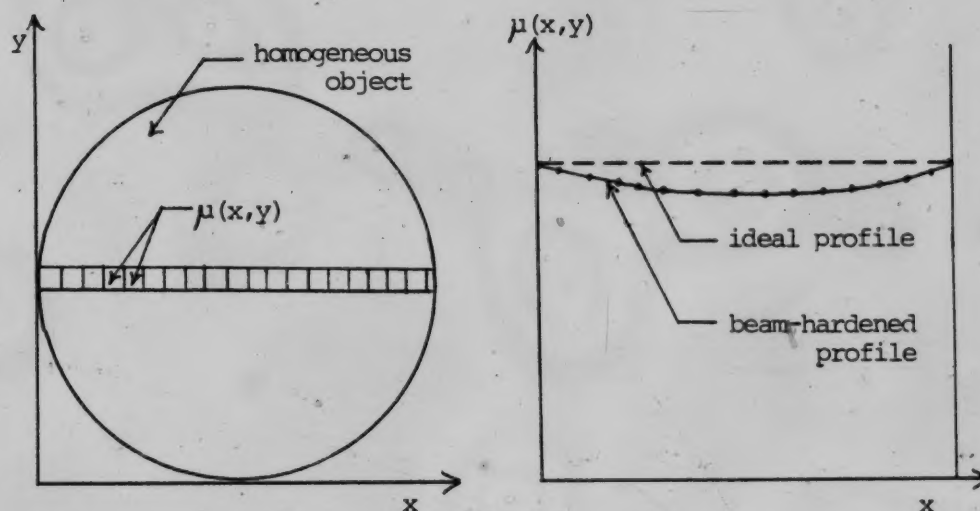


Figure 22. Projection Profile of a Reconstructed Image: The Cupping Effect.

CT numbers for homogeneous materials will have decreasing values towards their centers.

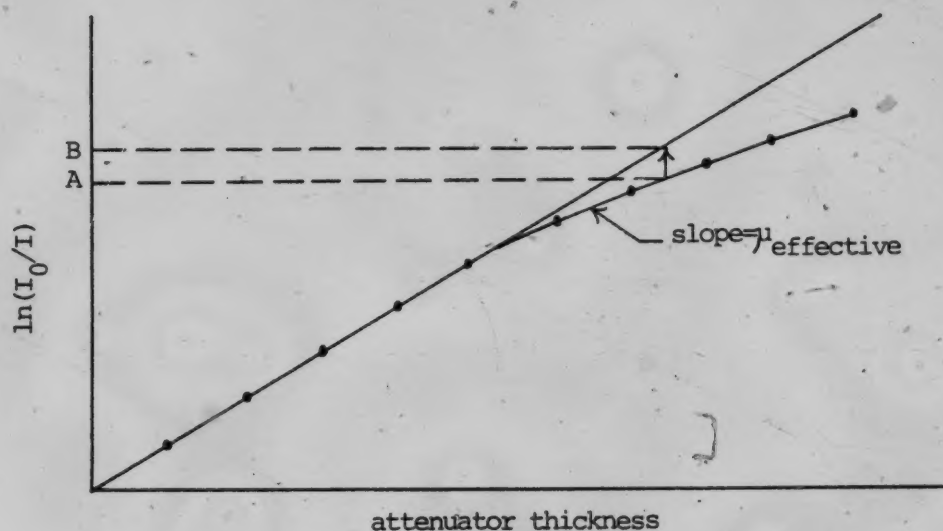


Figure 23. Beam-Hardening Correction.

Point A represents an uncorrected projection value; point B the corrected projection value, based on the beam-hardening curve measured with "sandwiched" plates of 1.65 mm thick Al and 6.0 mm thick plexiglass.

curves (2nd order polynomials) were fitted for each detector (Figure 23). In this manner, the deviation from linearity for a given projection value can be determined, and the data can be appropriately corrected. This technique of curve fitting is the same procedure as used for the dead-time corrections. A perfect correction can be made using this method for an object consisting of a homogeneous material. However, applying this type of beam-hardening correction to a multi-material structure such as the human forearm causes high density materials like cortical bone to have lower than expected CT numbers (an example of this type of beam-hardening artifact is shown in Figure 36). Errors in corrected projection values for soft tissue are negligible because when I approaches I_0 , the beam-hardening curve deviates little from the straight line. The beam-hardening procedure was tested by plotting the corrected projection values versus attenuator (plexiglass-aluminum) thickness. A linear correlation coefficient of better than 0.990 was attained for all detector units ($p < 0.0001$). However, streaking artifacts are still going to occur due to beam-hardening artifacts at high density edges.

A better method to produce accurate CT numbers, but one which requires longer computational time, is a material selective, iterative approach [32]. Accuracy is an important factor for comparison of data between research laboratories and between patients in a given study group. However, the primary purpose of this CT scanner is to produce bone

density values with high precision so that small density changes can be monitored over the course of time. Hence, for this application, accuracy is of secondary importance to precision, and the aluminum-plexiglass correction procedure was considered to be adequate.

4.6 Centering Fan-Beam

It is important that the correct offset for the front plate (which supports the source and detector assembly) be determined before detector angles and SCR distance are calculated. The reconstruction program assumes that in the scanner's "home" or reset position, corresponding detectors on opposite sides of the fan beam have equal (r, θ) -values relative to the SCR line, where r is the perpendicular distance from detector beam to CR, and θ is the angle from detector beam to SCR axis.

The process by which a source/detector assembly is positioned at the CR can be explained by first considering a one detector assembly (Figure 24). Count-rates are determined at each source position along the translational path length. The point at which the source/detector assembly passes through the wire is recorded as a drop in count-rate: x_1 can therefore be determined. The scanner assembly is rotated 180° and translated through the wire again to determine x_2 . If the center of the pathlength corresponds to the CR of the scanner, $x_1 = x_2$. Otherwise the start position must be appropriately moved a distance $\Delta x = (x_1 - x_2)/2$ in

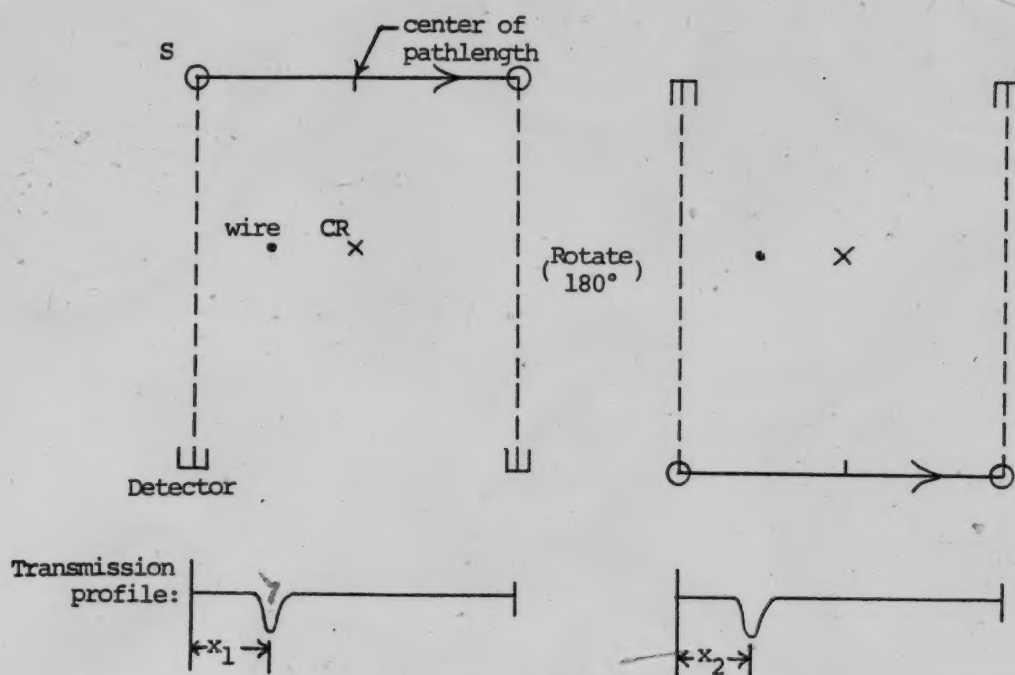


Figure 24. Method to Center Fan Beam.

If center of translational path length corresponds to CR, $x_1 = x_2$; otherwise move start position by an amount $(x_1 - x_2)/2$.

order to center the source/detector assembly.

This principle can easily be extended to a multi-detector fan beam assembly. In this case, x_1 and x_2 are determined for each detector in the fan beam and Δx is calculated by the following formula:

$$\Delta x = \left(\sum_{n=1}^N x_{n1} - \sum_{n=1}^N x_{n2} \right) / 2N \quad (15)$$

where N is the total number of detectors passing through the

wire.

It is also important to determine Δx for a number of SCR distances. Δx should be independent of SCR distance, otherwise the center line of the fan beam is not parallel to the y-movement, which changes the SCR distance. The detector assembly would need to be moved with respect to the source in order to correct for this misalignment.

Finally, it was also noted that Δx varies as the X-ray tube anode heats up. In other words, the source focal spot shifts as the anode heats up (see section 4.8); hence it is necessary to allow the tube to stabilize (about 25 min.) before centering the fan beam.

4.7 Geometric Parameters

Other geometric parameters of the fan beam that must be established with high accuracy are SCR distance and detector angles about SCR axis. The problem here is that the anode focal spot cannot be directly accessed, so any parameters which involve its position must be determined indirectly. In addition, detector angles are dependent on the source/detector distance and on the position of the anode focal spot.

The procedure to determine these two parameters is to place a wire outside the fan beam and rotate the scanner 360° such that the wire passes through each detector beam twice - at points A_i and B_i for the i-th detector (Figure 25). This procedure assumes that the fan beam has

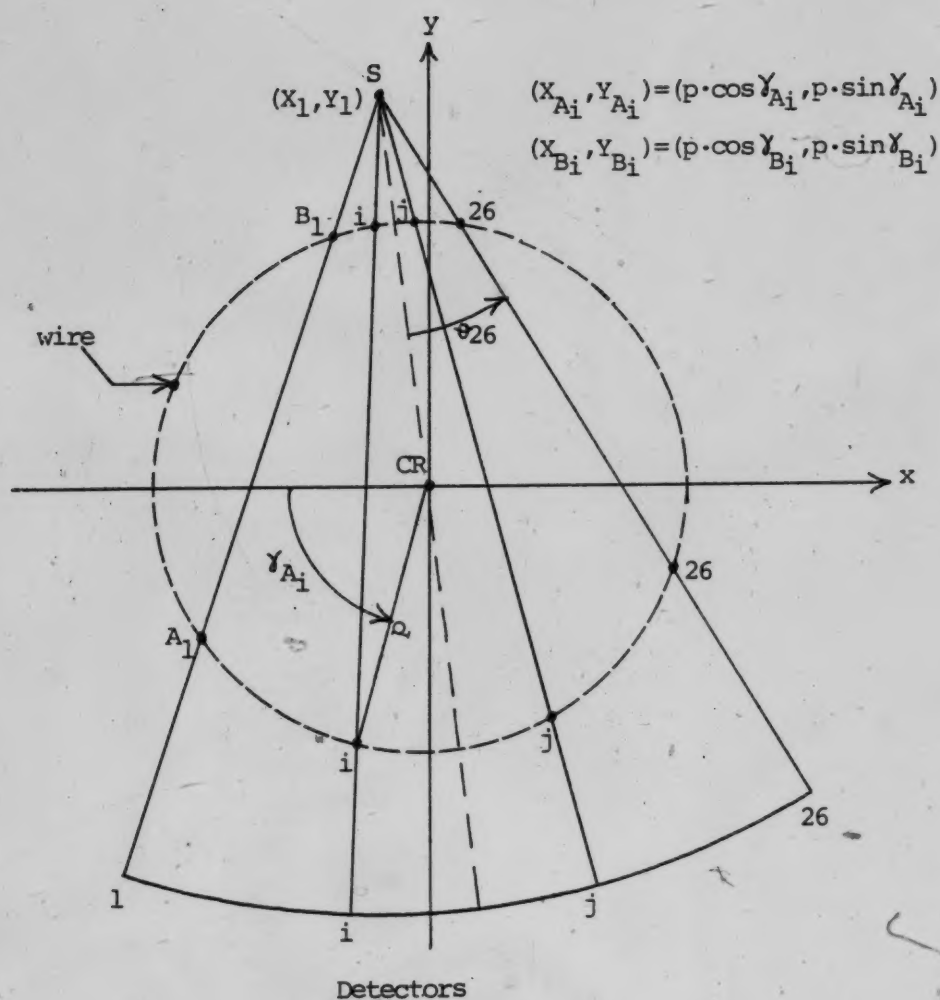


Figure 25. Fan-Beam Parameters.

A 360° rotation of the fan-beam through a wire determines points A and B for each detector unit. From these values detector angles θ and SCR distance are determined.

already been correctly centered. The coordinates for the points A and B depend on the CR-to-wire distance, p , and on the scanner rotation angle γ ; that is, the coordinate

$(x_1, y_1) = (p \cos \gamma_1, p \sin \gamma_1)$. Detector angles (relative to the

SCR axis) are calculated by determining the slope of each of the detector beams, and therefore do not depend on the parameter p . That is

$$\theta_i = \arctan(1/m_i) - \arctan(1/m_0) \quad (18)$$

where m_0 is the slope of the SCR axis and m_i is the slope of the i -th detector beam. The slope of the SCR axis is calculated by estimating the intersection point, S , for all lines AB (CR is the origin of the x,y coordinate system).

Detector angles, calculated using this experimental method, contain random errors. To determine the A and B values with high precision, the scanner motion description parameter $SINT$ must be made very small. The I/O buffer for the VAX is limited to 64 Kbytes, so the number of data points collected for any one rotation is restricted;

$$NUM\ DETECTORS \times NUM\ INTERRUPTS \times 4\text{-Byte Data} \leq 64\ KBytes.$$

The only parameter which can be varied is the number of detectors. Hence, instead of one rotation utilizing all 26 detectors, five rotations are necessary (with a different group of 6 detectors collecting data at each rotation) to determine the A and B values. Consequently, detector angles contain random errors. In order to determine accurate detector angles for the reconstruction program, the following procedure was applied:

1. The theoretical positions of focal spot (source/detector distance of 56 cm, source centered above detector

assembly) and detector collimators (precise machining - 2° apart at a source/detector distance of 56 cm) were assumed.

2. The total error between theoretical and experimental detector angles was calculated.
3. The theoretical focal spot was moved by increments in the x and y-direction.
4. The errors were re-calculated.
5. The location of the theoretical focal spot with the minimum error was chosen.

The source position was found to be 56.1 cm above and 2 mm to the left of the center of the detector assembly. The theoretically determined angles, based on this source position, are found in Appendix II.

To determine the SCR distance, the actual coordinates of the intersection point S (which depends on the parameter p) must be estimated. That is

$$\begin{aligned} \text{SCR}_1 &= ((X_1-0)^2 + (Y_1-0)^2)^{1/2} \\ &\propto p \\ &= k_1 p \end{aligned} \tag{16}$$

where k_1 is a proportionality constant. The distance p cannot be directly measured; to calculate its value, the SCR distance was incremented by an amount d, and the data collection procedure repeated to determine a second proportionality constant, k_2 . Assuming that the vertical

movement of the scanner is exactly aligned with the SCR axis, the following equality is derived:

$$\begin{aligned} d &= SCR_2 - SCR_1 = k_2 p - k_1 p \\ &= (k_2 - k_1) \cdot p \\ \rightarrow p &= d / (k_2 - k_1) \end{aligned} \quad (17)$$

Hence, a physical value for SCR is determined as, $SCR_1 = k_1 \cdot d / (k_2 - k_1)$. This procedure was repeated on several occasions and a first estimate of SCR distance (relative to the vertical movement position switch) was determined to a precision of ± 3 mm. A more refined estimate of SCR was achieved by determining the extent of the artifacts in a series of reconstructed images as the SCR parameter was varied (section 6.3 describes the method by which the severity of the artifacts is examined).

4.8 Thermal Characteristics of Anode

A final consideration for the calibration procedure is the manner in which the count-rate for each detector varies during anode warm-up/cool-down. In performing a multiple-rotation scan, X-rays reaching detectors 1 and 26 never pass through the object; these detectors always measure open count-rates. Hence, any electronic drift over the course of a scan is accounted for in the reconstruction process by monitoring the open-count values of detectors 1 and 26. That is, it does not matter if the photon flux

increases proportionately for each detector during a scan, because detectors 1 and 26 monitor these changes. However, problems do occur when the open count-rates for individual detector units change in a disproportionate manner, such as when the X-ray tube is warming up.

Warm-up curves were established for each detector unit by measuring open-count rates at various times over the course of one hour. A mean count-rate was established at $t=60$ minutes for each detector unit, and offset values from this mean were calculated for the various measurement points (Figure 26). The anode is defined as being "stabilized" at the time when the variation in the range of offset values is less than one percent. Hence the warm-up period for the anode (from a cold start) is 25 minutes.

To circumvent count-rate errors which would occur for a scan where the anode has not been warmed up for 25 minutes, a calibration curve describing how count-rate varies for each detector over the first 25 minutes was produced. The warm-up curves for each detector unit were determined on several occasions. These curves are established in the same manner as the dead-time and beam-hardening curves, and found to be very consistent over time. The warm-up curves also show that count-rates increase as stabilization is reached. This leads to the speculation that as the anode warms up and expands, the source/detector distance is decreased by approximately 0.3 mm (based on the inverse-square law) to account for the 1% increase in count-rates over the first 25

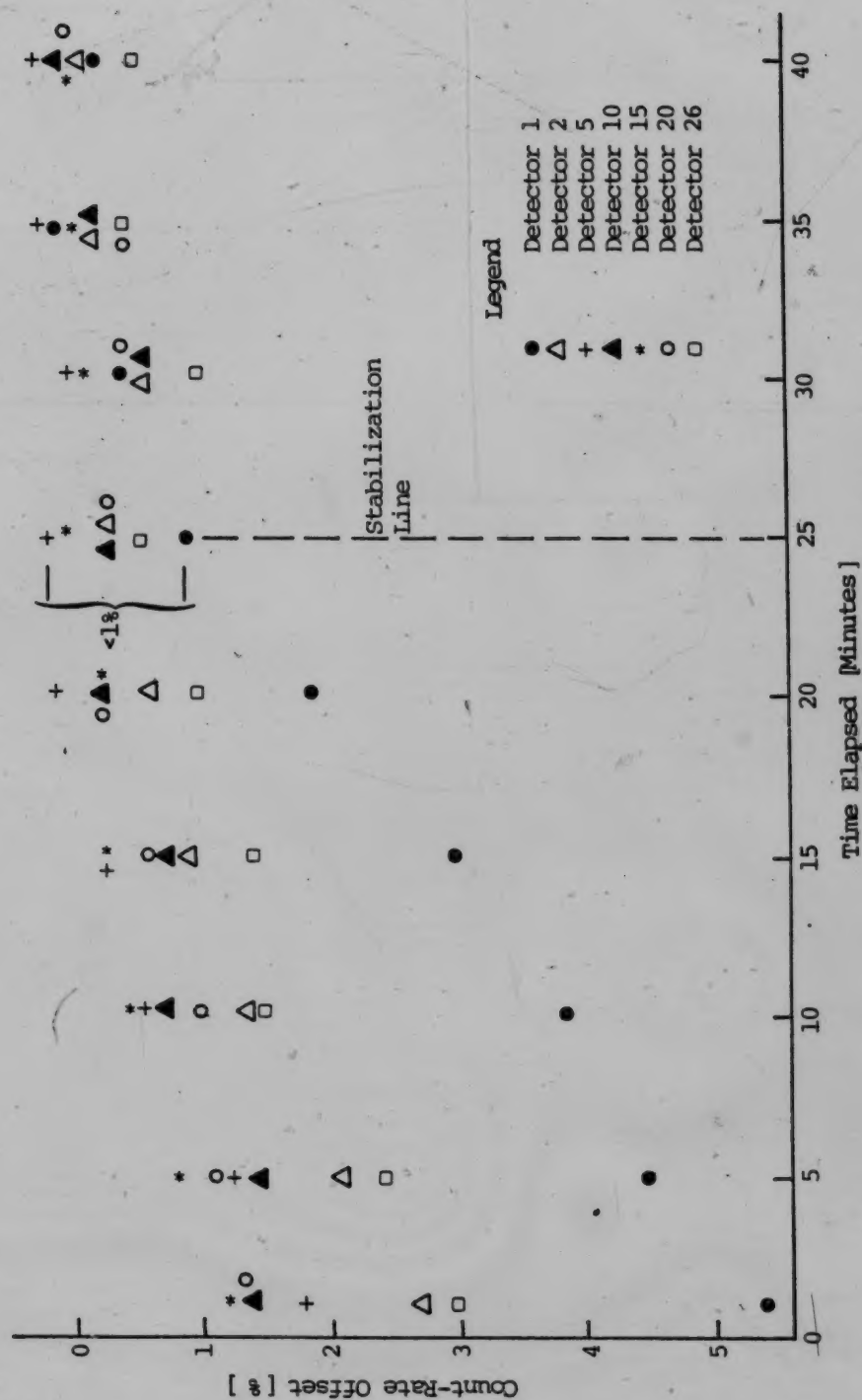


Figure 26. Detector Warm-up Curves (from cold start).

The detection system is considered stable at $t=25$ min: the point at which the range of offset values is less than 1%.

minutes for the center detectors. The HV generator for the X-ray tube is very stable over time and temperature ($10^{-3}/^{\circ}\text{C}$) [21] and cannot account for this change. Increases in count-rates for the outer detectors are more the result of better collimator alignment than of the inverse-square law.

The normal operational procedure for the scanner would be to turn the tube down or off between scanning patients in order to prolong tube life. Hence, it is also necessary to know the cool-down characteristics of the tube. That is, if the tube is turned off for 20 minutes, how long is the warm-up period of the tube before stabilization is once again achieved? These curves were established for three cases: (1) X-ray tube turned off for various time periods, (2) X-ray tube turned down to 5.0 kVp, 0.0 mA (this is the lowest tube setting without turning the tube off), and (3) X-ray tube turned down to 25 kVp, 1.0 mA (Figure 27). Count-rates at the last tube setting are no higher than background count-rates. By simply monitoring the cool-down period of the tube, a corresponding warm-up period can be established and count-rates adjusted accordingly. In this manner, images can be produced immediately after the tube is turned on.

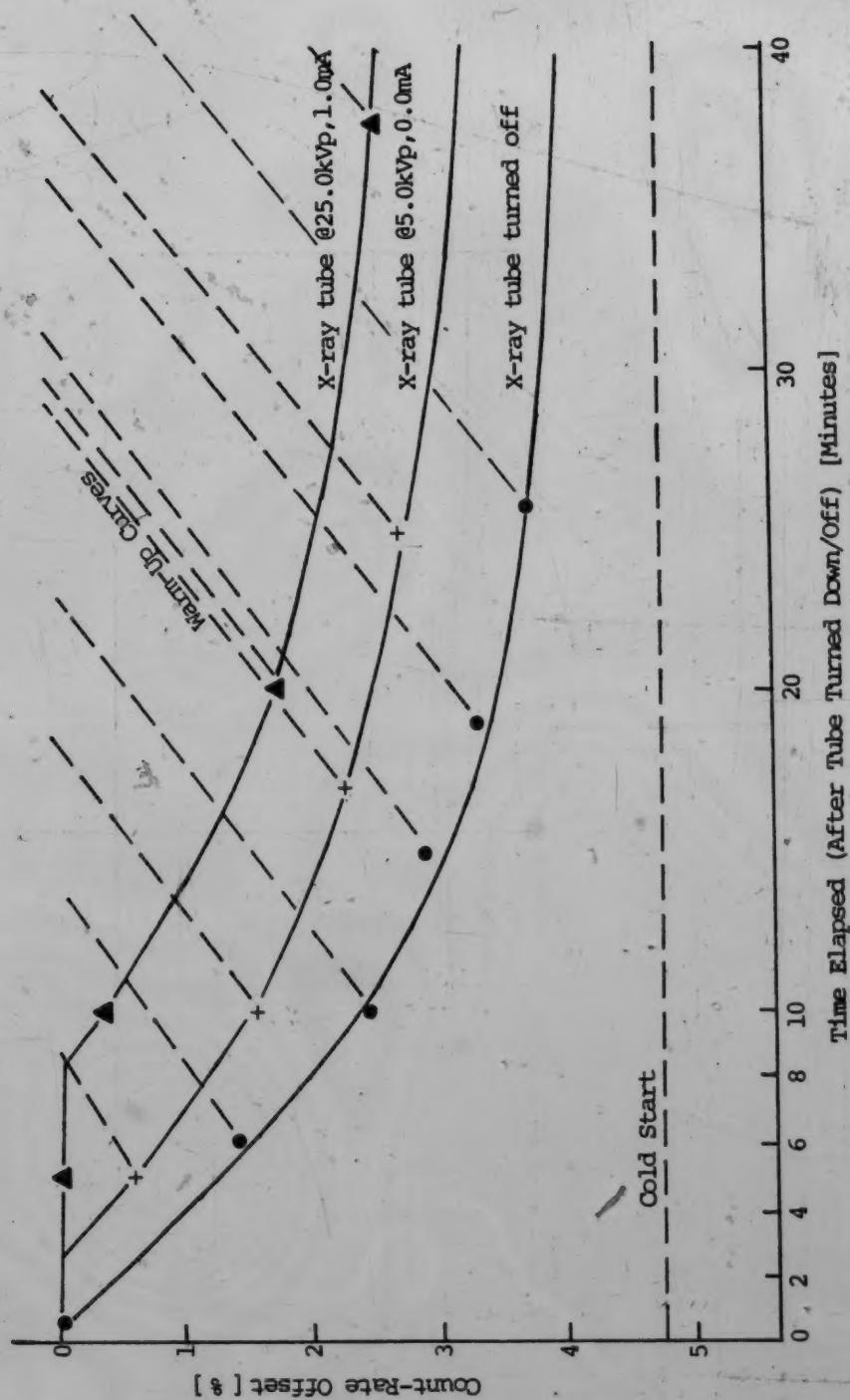


Figure 27. Detector 1 Cool-Down Curves.

The cool-down response for other detector units is slower.

5. Data Collection

The method by which data are collected to produce a reconstructed image of high spatial resolution is unique to this CT scanner. The scanner collects data in the rotate-only mode; however, with only 26 detector units, one rotation does not produce enough data to achieve the necessary high spatial resolution. To avoid this problem, multiple rotations are performed with each rotation separated by a small translational movement to simulate a new set of detectors. Hence for five rotations, $5 \times 26 = 130$ detector units are simulated, providing enough data points or projection values to fill a 256×256 projection matrix.

Figure 28 illustrates the geometry of the projection measurement process (where only 4 detectors are shown for clarity of presentation). An X-ray beam traverses the object along the line $L(r, \phi)$ and its intensity $I(r, \phi)$ is measured by the detector. If the function of the object, $f(x, y)$, represents the linear attenuation coefficient at point (x, y) , then a projection value $p(r, \phi)$ is defined as

$$p(r, \phi) \equiv \ln(I_0/I(r, \phi)) = \int_{L(r, \phi)} f(x, y) dl \quad (19)$$

where I_0 is the intensity in air. In this illustration, 4 projection values (one per detector) are determined at each source position.

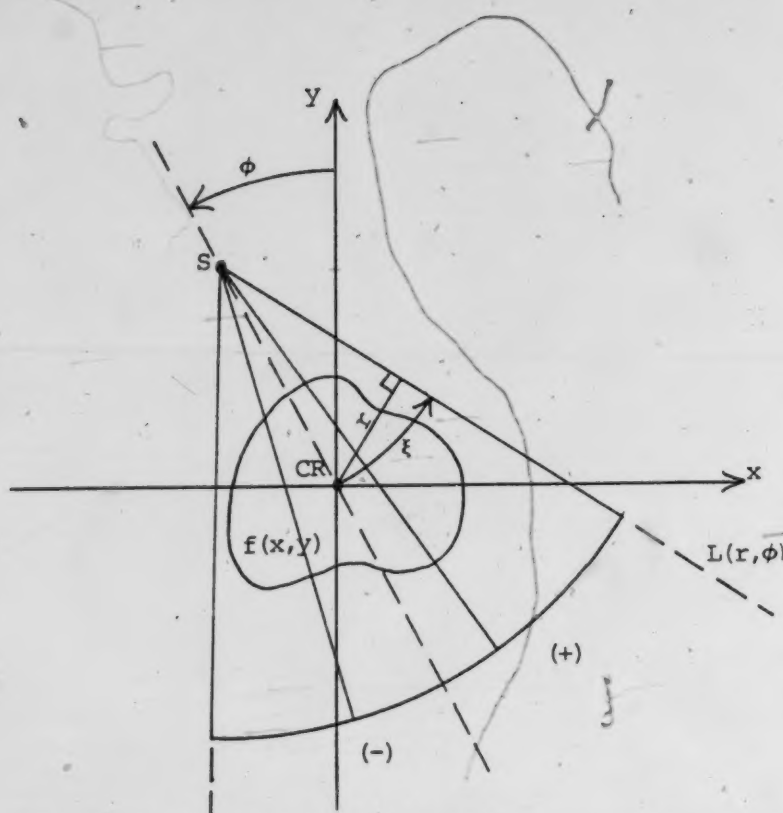


Figure 28. Geometry of the Projection Measurement Process.

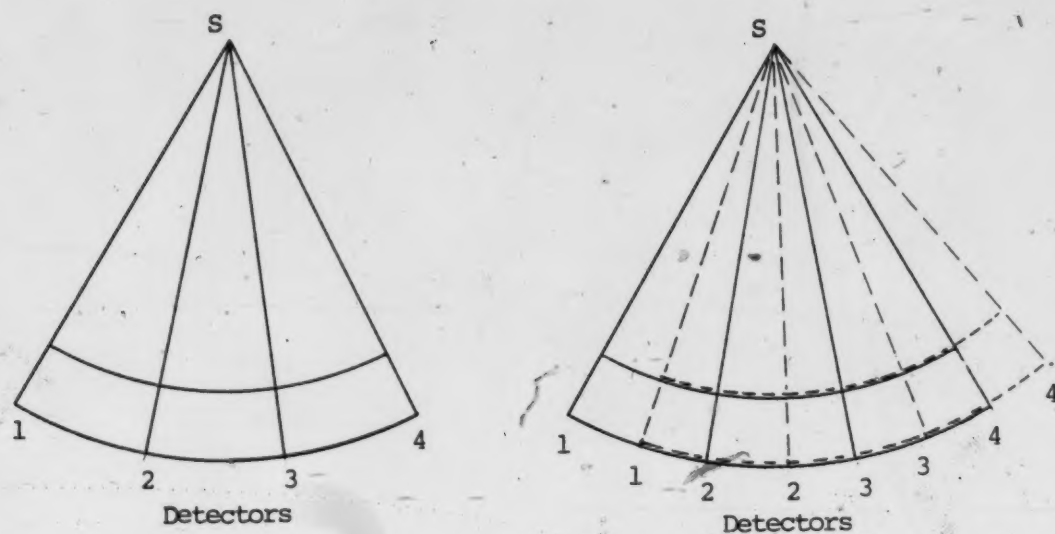
In this illustration, four projection values are determined for each source position as the fan beam rotates about CR. The r -coordinate for detectors on the right-hand side of the SCR axis are arbitrarily labelled positive, and on the left-hand side are labelled negative. ξ is the arclength from the CR to the particular detector beam.

All scanners measure a set of projections and from these data, the "density" distribution $f(x,y)$ is computed using some reconstruction algorithm.

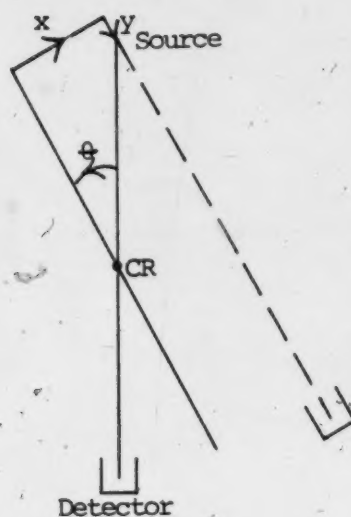
5.1 Rotation Scheme

The method of improving spatial resolution by offsetting the fan beam detector from its symmetrical position by a fraction of the detector spacing and by making

measurements over 360° has been described by Peters and Lewitt [33]. The offset is implemented by rotating the detectors about the source so that the radial and angular coordinates for each detector ray are changed (Figure 29a). A direct rotation about the source is not possible with the gantry system of this scanner. Instead a motion involving three of the scanner movements is used to simulate a rotation about the source (Figure 29b). Each rotational offset about the source creates a new set of detector positions, hence 5 offsets simulate 130 detector positions. If the detector fan was symmetric about the SCR axis, then half the data collected over 360° would be redundant (i.e. $p(r_1, \phi_1) = p(-r_2, \phi_1 + 180^\circ)$ when $r_1 = -r_2$). An initial rotational fan offset alters the symmetry about the SCR axis so that the data collected from detectors with opposing views interleave instead of overlap in the radial coordinates of the projection plane. Hence a total of $130 \times 2 = 260$ radial coordinates are determined for each 5-rotation scan. By varying the number of detector offsets per scan, the spatial resolution can be altered. The advantages of decreasing the number of detector offsets or radial coordinates in the projection plane are shorter scan times and decreased dose per scan, at a cost of decreased spatial resolution.



(a)



$$x = SCR \sin(\theta)$$

$$y = SCR (1 - \cos(\theta))$$

(b)

Figure 29. Rotational Movement of Detectors About Source.

(a) Source is center of rotation, (b) rotation about the center of the projection plane and translations in the x- and y-direction.

5.2 (ξ, ϕ)-Projection Plane

The manner in which data are placed into the projection plane is shown in Figure 30a. The parameter ξ is used instead of r for ease of illustration; that is, the r -spacing between adjacent detectors is not equal, whereas the arclength spacing, ξ , is. Each diagonal connects the projection values collected for one source position. With this fan beam geometry, 26 projection values are equally spaced along each diagonal in the projection plane (only 4 detectors are shown). The spacing between parallel diagonals, i.e. the angular increment of the source position, is equal to the detector spacing so that data points in the projection plane are also aligned vertically. Data are collected every $2/3^\circ$, so over a 360° rotation, 540 angular coordinates and only 26 radial coordinates are determined. A small rotation about the source shifts the detector coordinates along the diagonal (Figure 30b). Variation of the number of rotational offsets about the source dictates how many detector positions are found on each diagonal: 5 rotational offsets result in 130 data points per diagonal. The 5 rotational offsets are carefully chosen so that the order of the detectors is not changed; that is, the sum of the 5 offsets is not greater than the spacing between detectors. The data is interleaved (Figure 30c) to give 260 radial coordinates per scan. In general, given an N -detector assembly, $2 \cdot N \cdot n$ radial coordinates are simulated, where n is the number of

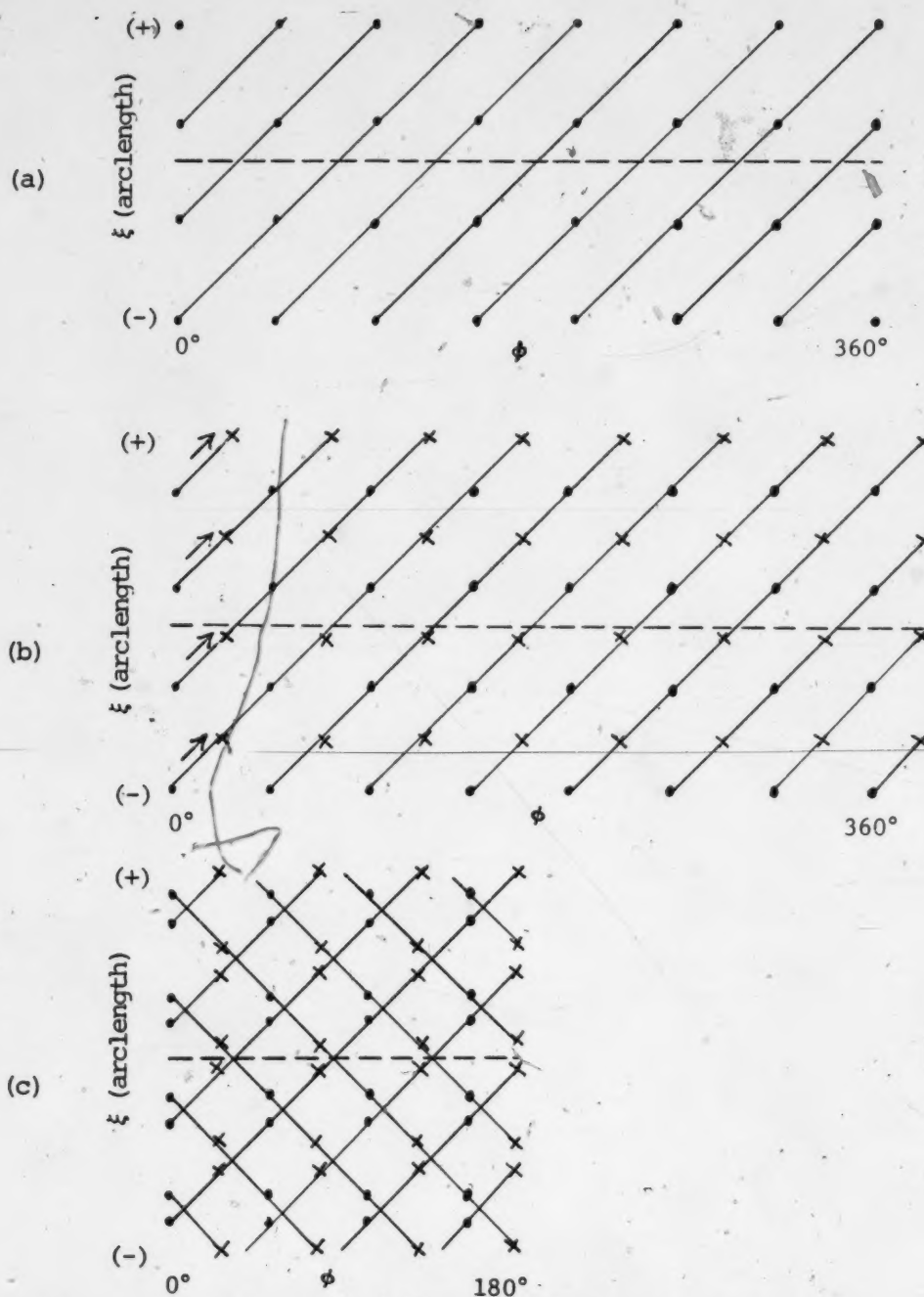


Figure 30. (ξ, ϕ) -Projection Plane: Rotational Format 1.

Data points in projection plane where points collected simultaneously are connected (a) after 1 rotation, only 4 detectors are shown, (b) after 1 rotation about the source and a second 360° rotation, (c) after interleaving data. The dashed line represents the r -coordinate for CR.

rotational offsets per scan. Each rotation about the source produces another set of angular coordinates as well as radial coordinates. In addition, the arclength spacing between ξ -coordinates is equal, however the translational spacing, r , in millimeters is not. Therefore this type of rotational scheme requires interpolation in both the radial and angular coordinates before convolution and backprojection can be carried out.

The geometry of this scanner suggests the use of a direct fan beam reconstruction algorithm, however with a multiple rotation data collection procedure, no reconstruction algorithm can be utilized until data from the last rotation are collected. It was therefore decided [34] to use the faster parallel beam reconstruction algorithm [35,36] along with a special scheme of re-ordering the data. In order to simplify this "re-binning" scheme, the 5 small rotations about the source have been replaced with 5 translations in the x-direction so that no interpolation in the projection plane's angular coordinate is necessary (Figure 31). As a consequence, geometrical resolution is increased with this rotation scheme and computing time reduced.

This rotational scheme therefore leads to an equivalent parallel ray data set in projection space consisting of 260 radial coordinates and 270 angular coordinates. For a data matrix consisting of N radial detector positions, the total number of pixels (A) within the circle of reconstruction is

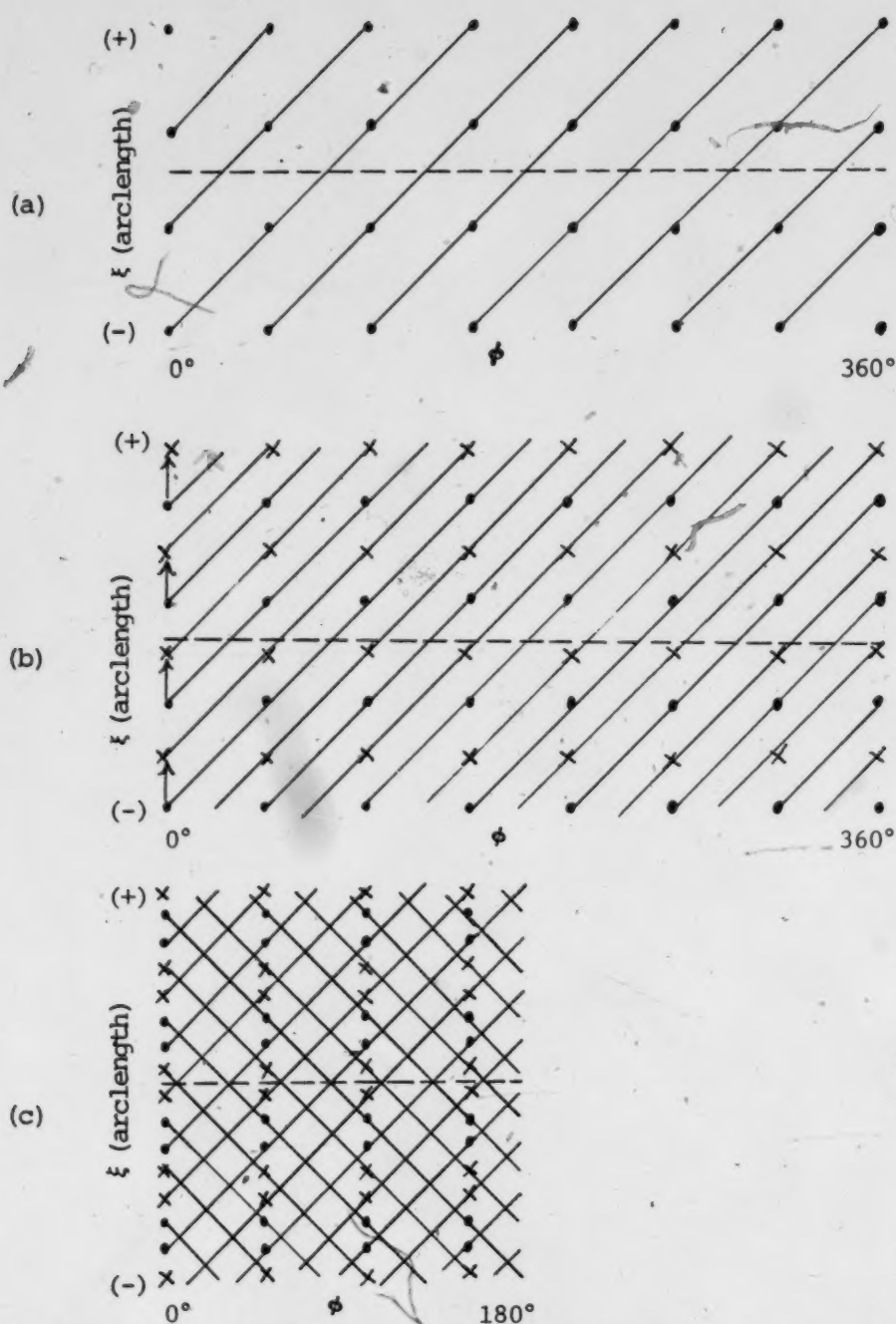


Figure 31. (ξ, ϕ) -Projection Plane: Rotational Format 2.

Data points in projection plane where points collected simultaneously are connected (a) after 1 rotation, only 4 detectors are shown, (b) after translation in the x-direction, and a second 360° rotation, (c) after interleaving data.

$$A = \pi \cdot (N/2)^2$$

For an accurate determination of the pixel values, the total number of measured projection points must at least be equal to the number of reconstructed pixel values. Hence the number of angular views (M) over 180° for a projection plane with parallel-beam ordering is

$$M = A/N = \pi N/4 \quad (20)$$

If M is significantly smaller than $\pi N/4$, the numerical approximations of the filtered backprojection method result in reconstruction errors [37]. For M greater than $\pi N/4$, the effects of projection measurement noise are reduced by averaging together projection values. For the described data collection scheme, where $N=260$, M would be 202 according to equation 20. By actually collecting data for $M=270$ angular views, statistical noise in the reconstructed images is reduced.

The spacing between data points in the radial coordinate of the projection plane is equalized by linear interpolation, after which standard convolution and backprojection techniques are carried out. The reconstruction procedure has the following mathematical form:

$$f(x,y) = 2(N/M) \sum_{\phi} \sum_r p(r,\phi) * g(r-r') \quad (21)$$

$$\begin{aligned} g(r-r') &= 1 \\ &= -1/(4|r-r'|^2-1) \quad , \quad r=r' \\ &\quad , \quad r \neq r' \end{aligned}$$

where $g(r-r')$ is a 256-point numerical version of the Shepp-Logan convolution kernel [38].

6. Evaluation of the CT System

The new CT scanner has been developed to measure trabecular bone density with high precision in serial studies. Any artifact in the reconstructed images degrades the precision of the bone evaluation procedure. This section examines the various types of artifacts which can arise. The causes of these artifacts are three-fold: (1) the unique data collection scheme of this CT system, (2) physical parameters of the system which cannot be corrected for (e.g. finite beam-width, polyenergetic source), and (3) calibration errors. The precision and tolerance (with respect to producing artifacts in the reconstructed images) of each calibration procedure is determined, and the system stability is analyzed. In addition, the capabilities of the system are illustrated through some measurements on patients.

6.1 Spatial Resolution

Spatial resolution is one parameter which can be used to analyze the capabilities of a CT system. A high spatial resolution means edges in a reconstructed image can be defined very sharply. With respect to bone density evaluation, the inner 40-60% area of the bone cross-section (the trabecular bone region where density analysis is done) is defined from the outer edge of the cortical bone. The spatial resolution of any CT system is influenced by a number of factors:

1. Pixel size
2. Beam width
3. Sampling frequency in spatial domain
4. Contrast of structures to be identified
5. Source intensity or measurement time
6. Convolution kernel

6.1.1 Pixel Size

Pixel size ultimately dictates spatial resolution for any display system. This parameter is determined by the physical dimensions of the reconstruction plane. The size of the plane is dictated by the distance of the outer detector beams from CR and varies with SCR distance. Hence for a 256x256 reconstruction matrix, pixel size ranges from 0.345 mm for small objects (75 mm diameter) to 0.855 mm for large objects (220 mm diameter).

6.1.2 Projection Beam Width

In practical CT work, the projection beam can no longer be considered as a line from source to detector. Instead beam width should be thought of as the FWHM of the intensity profile of photons passing through the collimator slot (Figure 32) [39]. Hence, the photon beam width is a function of source focal width, collimator width, source/detector distance, and SCR distance. The photon beam width has a range from 0.696 mm (large objects) to 1.056 mm (small

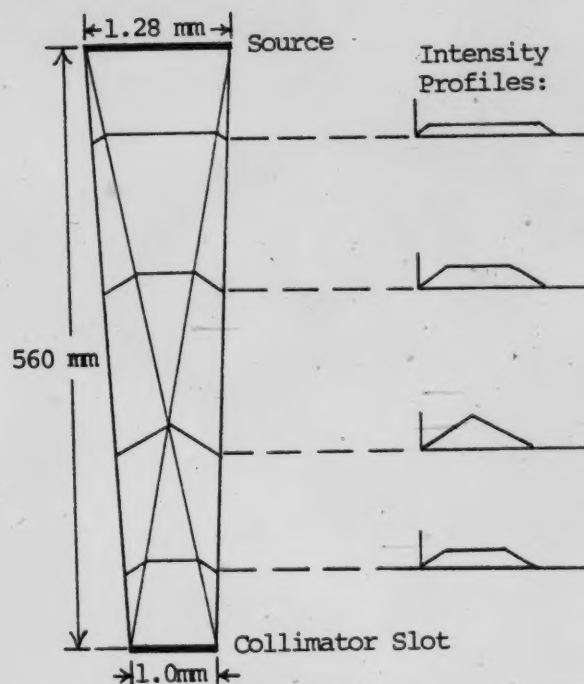


Figure 32. Beam Width Intensity Profiles.

Beam width is taken as the FWHM of the intensity profile at a given SCR distance.

objects). This value is the photon beam width without scanner movement. However, projection values are collected every $2/3^\circ$ or SINT=33.86 motor steps for this particular data collection scheme, so an arclength component must be added to determine the actual projection beam width. This arclength value is calculated at one half the object radius:

$$\text{arclength} = \frac{2\pi \cdot (2/3^\circ)}{360^\circ} \cdot (\text{Object Radius}/2) \quad (22)$$

Arclength has a range of values from 0.225 mm for small objects to 0.625 mm for large objects.

Therefore, the projection beam width, which equals photon beam width plus arclength, has a range of values between 1.28 and 1.32 mm. Geometrical resolution is limited to one half the projection beam width; that is, the smallest object diameter which can be "resolved" is 0.64 mm.

6.1.3 Spatial Sampling Frequency

A third factor which influences geometrical resolution is the spatial sampling frequency. A repeating pattern of points can only be correctly distinguished from the medium if the sampling frequency of the scanner is at least twice as large as the spatial frequency of the points (the Nyquist criterion [40]).

The angular sampling frequency for this scanner is determined by the average arclength of the projection interval which has a range of values from 0.225 mm to 0.625 mm for a measurement interval of $2/3^\circ$ or $M=270$. The radial sampling frequency depends on the diameter of the projection plane and on the number of detector offsets; the distance between points has a range from 0.32 mm to 0.79 mm, for $N=260$. The average sampling area for a 75 mm object is $0.32 \text{ mm} \times 0.225 \text{ mm} = 0.072 \text{ mm}^2$, and for a 220 mm object equals 0.49 mm^2 . Hence the smallest object size which can be resolved, assuming an infinitely small projection beam width, has a range of diameters from approximately $(0.072 \text{ mm}^2)^{1/2} = 0.27 \text{ mm}$ (small objects) to 0.71 mm (large objects).

6.1.4 Contrast

Another factor affecting spatial resolution is the density contrast. The spatial resolution for elements with high density difference or contrast, such as bone/tissue interfaces, will be better than low contrast elements such as fat/tissue interfaces because statistical noise in the reconstructed images obscures small low contrast elements. The number of counts recorded in a given interval will vary due to the Poisson distribution of photons emitted from the anode. If the statistical fluctuations in count-rate are large compared to the count-rate differences for the two materials, then a given sequence of repeating elements will be lost in the noise of the reconstructed image.

6.1.5 Photon Statistics

As already mentioned, one possibility for distinguishing the elements in a repeating sequence is to increase the contrast between elements. Another possibility would be to increase the measurement time or source intensity, thereby reducing the statistical noise in the reconstructed image. Increasing the number of counts recorded per measurement interval (and the patient dose) by a factor of N , decreases the relative error in the projection counts by $1/\sqrt{N}$.

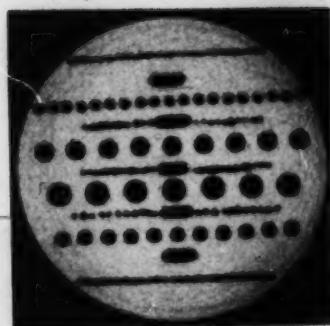
6.1.6 Convolution Kernel

The final factor affecting spatial resolution is the shape of the convolution kernel. The higher the frequency response of the convolution kernel, the greater the resultant spatial resolution; however, this tends to increase noise in the reconstructed image [41]. The sharpness of the 256-point convolution kernel used with this CT system will vary depending on the diameter of the projection or reconstruction plane, hence the frequency response of the kernel will increase for smaller objects.

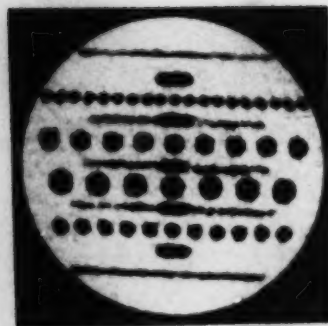
6.1.7 Measurement of Spatial Resolution

All six of the discussed factors have some bearing on the spatial resolution of a system. The projection beam width for this new CT scanner has been shown to limit the spatial resolution to object diameters of 0.64 mm. This resolution might be further degraded by such factors as object contrast, measurement time, or the frequency response of the convolution kernel.

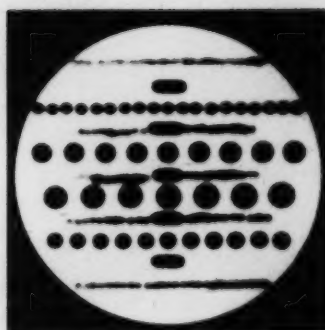
A plexiglass phantom (Figure 33) was used to compare the high contrast spatial resolution of the new CT scanner to an ¹²⁵I scanner developed at the University of Alberta [14], and two commercial systems - a Picker Intl 1200SX, and a GE 9800 CT/T scanner (Table 5). The phantom contains a series of holes ranging in size between 0.5 mm and 4.0 mm diameter and the plexiglass spacing between holes equals the hole diameter. The spatial resolution values



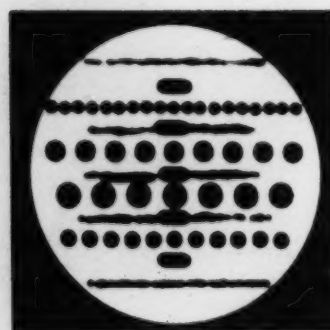
(a)



(b)



(c)



(d)

Figure 33. Comparison of Spatial Resolution.

(a) Specialized X-ray scanner, (b) ^{125}I scanner, (c) Picker Intl 1200 SX, (d) GE 9800 CT/T. The hole diameters from top to bottom in the reconstruction phantom are 0.78, 2.3, 0.78, 3.8, 0.52, 3.2, 0.68, 1.46, and 0.52 mm. In each case, the spaces between holes are equal to the hole diameter. Unfortunately, due to mechanical drilling problems, the small diameter holes (0.52 and 0.68 mm) are not all equidistant.

Table 5. Comparison of Spatial Resolution.

CT Scanner	Scan Parameters	High Contrast Spatial Resolution (mm)
New X-ray scanner	50kVp, 2.8mA, 20min	0.6
¹²⁵ I scanner	30keV, 30min	1.1
Pickler Intl 1200SX	80kVp, 40mA, 2sec	0.8
GE 9800 CT/T	80kVp, 40mA, 2sec	1.0

Spatial resolution was defined as the smallest hole diameter which could be distinguished in the reconstructed image. (n.b. The spatial resolution values attained in this manner depend on the viewer, and so these numbers are only valid in a comparative sense).

quoted in Table 5 involved some visual interpolation. These results show that a 26 detector scanner using a multiple rotation scheme can achieve a spatial resolution comparable to or better than the spatial resolution of typical commercial whole body CT scanners utilizing far more detectors units.

6.2 Artifacts

The artifacts which appear in the reconstructed images of this scanner can be classified into two types:

- (1) artifacts due to geometrical, and counting errors;
- (2) artifacts produced by objects with high contrast interfaces or by objects with a large range of densities.

6.2.1 Type 1 Artifacts

Geometrical artifacts are a result of inaccurate calibration of the SCR distance, fan beam centering, or initial rotational fan offset. Translation of the source/detector assembly in the y-coordinate is used to maximize the number of detector beams passing through the object. Decreasing the SCR distance allows smaller objects to be scanned with the same relative geometrical resolution as larger objects. However, variation of the SCR distance causes the physical dimensions of the reconstruction plane to be altered; the size of the plane is dictated by the distance of the outer detector beams from the center. In order to place data points evenly over the diameter of the reconstruction plane, the SCR distance needs to be accurately determined. If the SCR distance is underestimated, the x-translational increment will cause a clustering of data points in the ξ -coordinate (the sum of the 5 detector offsets will be less than the distance between two detectors), whereas an overestimated SCR distance causes an incorrect ordering of the data points (Figure 34b).

Before the fan beam can be moved to its correct rotational offset position, it is necessary that the fan be properly centered, i.e. symmetrical about the SCR axis. A rotational offset of $1/4$ arclength of the spacing between detectors is implemented to properly interleave the data for opposing detectors in the fan. Errors in rotational offset

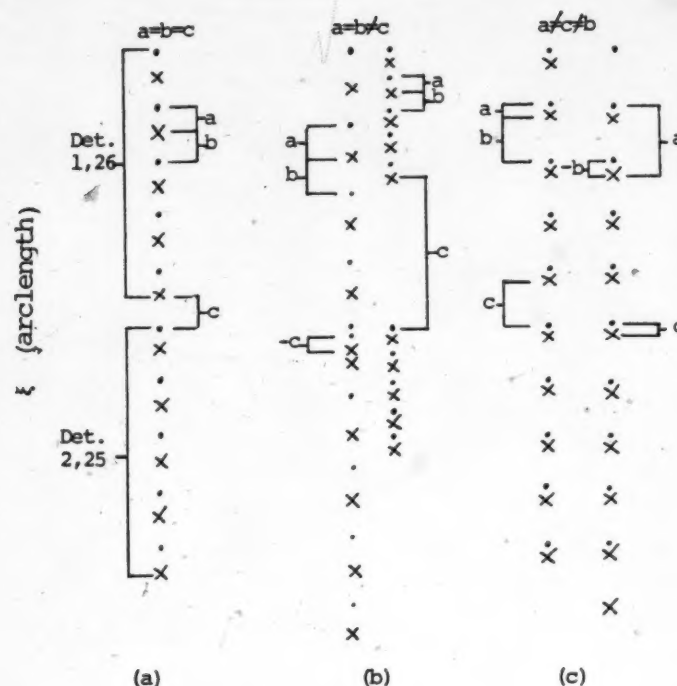


Figure 34. Effects of Geometrical Factors on Interleaving Process.

Interleaved data (at a particular ϕ -coordinate in the projection plane) for 2 pairs of detectors. a and b represent the arclength between adjacent data points for an interleaving detector pair, whereas c represents the arclength between the nearest data points of adjacent detector pairs. (a) Ideal interleaving, (b) variation of SCR distance, (c) variation of fan beam centering or rotational fan offset.

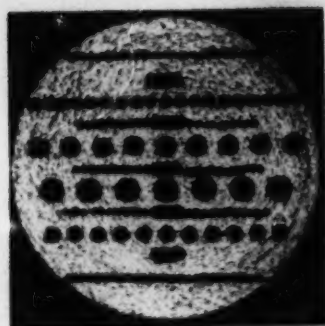
or fan beam centering cause the interleaved data to be incorrectly spaced or, with larger errors, to be incorrectly ordered (Figure 34c).

The influence of each of these factors on a reconstructed image of a plexiglass phantom is shown in Figure 35. The artifacts due to an erroneous value for the SCR distance are characterized by bands of concentric circles. The width of each band corresponds to the sum of

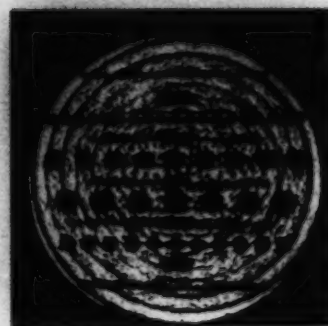
the 5 translational offsets (Figure 35b). Malfunctioning of a detector or insufficient accuracy in dead-time corrections, X-ray spectrum matching, or beam-hardening corrections show up as rings within one band for that detector (Figure 35c). An error of this nature also illustrates how the data are interleaved for corresponding detectors symmetrical to the center of the fan beam. If this interleaving process is not correctly implemented, as a result of erroneous fan beam centering or initial rotational fan offset, interpolation errors occur throughout the entire reconstruction plane (Figure 35d). In addition, fringes will appear at all contours.

6.2.2 Type 2 Artifacts

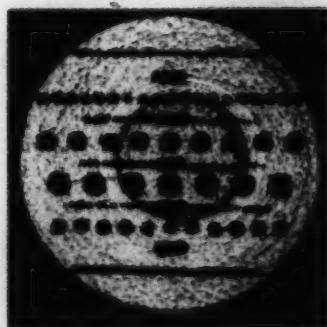
The second kind of artifact is due to unavoidable problems associated with scanning multi-density objects where there is a large range of densities. As previously mentioned, the beam-hardening correction assumes a constant mass attenuation coefficient throughout the object. For regions of the object where the mass attenuation coefficient is significantly greater than the assumed value, i.e. the value from which the beam-hardening curve is produced, reconstruction artifacts in the form of streaks will result. A plexiglass phantom with two aluminum cylinders to model the human forearm was used to illustrate these artifacts (Figure 36). Beam-hardening streaks (if visible) would show up as low density streaks between the aluminum rings of this



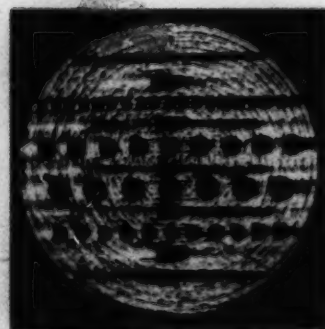
(a)



(b)



(c)



(d)

Figure 35. Geometrical and Counting Errors: Type 1 Artifacts.

(a) Plexiglass phantom with correct parameters (50 kVp, 2.8 mA, 0.7 mm Cu-filter, 1500 Hz); (b) SCR distance error of 2.5 mm; (c) dead-time error of 60% for detector 9; (d) fan beam centering error of 550 μm .

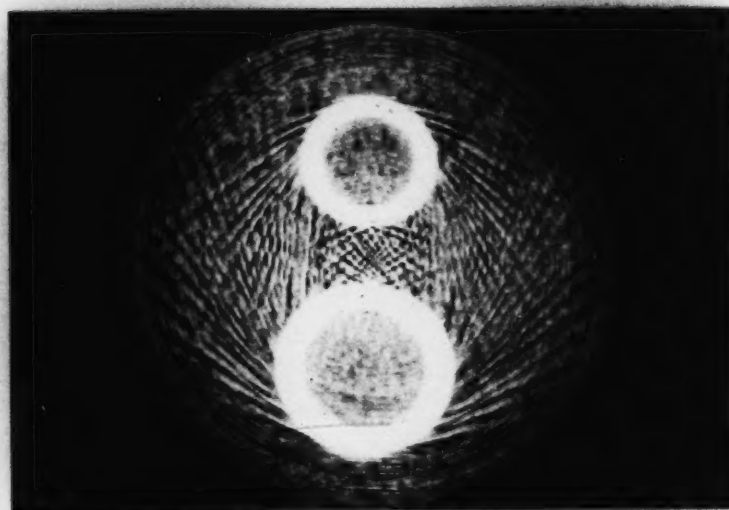


Figure 36. Aluminum Ring Phantom: Type 2 Artifacts.

Streaks between and around the outer edge of the aluminum rings is due to a combination of beam-hardening, aliasing, and exponential edge gradient effects.

object.

Artifacts also result from using a finite projection beam width. When a beam passes through an edge in the object, e.g. an aluminum/plexiglass interface, part of the beam passes through the higher density material and part through the lower density material, hence an average intensity is measured and a corresponding projection value calculated. However, this projection value, based on an average intensity, is underestimated; the correct projection value should be an average of the densities of the materials. The measured projection value is

$$p(r, \phi) = \ln[I_0 / (\alpha \cdot I_1(r, \phi) + (1-\alpha)I_2(r, \phi))] \quad (23)$$

where α is the fraction of one of the materials at a coordinate (r, ϕ) and $(1-\alpha)$ is the fraction of the other material. On the other hand, the projection value which represents an average of the material densities is,

$$\begin{aligned} p(r, \phi) &= \alpha \cdot p_1(r, \phi) + (1-\alpha)p_2(r, \phi) \\ &= \alpha \cdot \ln(I_0/I_1(r, \phi)) + (1-\alpha)\ln(I_0/I_2(r, \phi)) \\ &= \ln[I_0 / (I_1^\alpha(r, \phi) \cdot I_2^{1-\alpha}(r, \phi))] \end{aligned} \quad (24)$$

Equation 23 will be less than equation 24 if the following inequality is true:

$$\ln(\alpha I_1(r, \phi) + (1-\alpha)I_2(r, \phi)) > \ln(I_1^\alpha(r, \phi) \cdot I_2^{1-\alpha}(r, \phi)) \quad (25)$$

$$\Rightarrow \ln(\alpha I_1(r, \phi) + (1-\alpha)I_2(r, \phi)) > \alpha \ln I_1(r, \phi) + (1-\alpha) \ln I_2(r, \phi) \quad (26)$$

Joseph and Spital [42] have proven mathematically that this inequality exists; a graphical interpretation of this inequality is shown in Figure 37. This effect is called the exponential edge gradient effect (EEGE), and results in streaks in the reconstructed image. The streaks occur at edges, particularly at the outer edge of any material with circular geometry. The rotational data-collection scheme causes certain source/detector beams to "slide" along the edge of a circular object and thus amplifies the effect. If a square object was scanned, streaking artifacts due to the

density material in the neighborhood of a low density material, the number of counts recorded will be greater than if the low density material was removed. The increased counts are due to scattering and are dependent on the dimensions of the collimator. Artifacts due to scattering have been minimized by using detector collimators which are 5 cm deep x 1 mm x 3 mm to ensure that very few scattered or secondary photons reach the detectors.

6.3 Precision and Tolerance of Calibration Methods

Precision and tolerance levels need to be established for each of the calibration procedures. Ideally, the ratio of tolerance (with respect to producing artifacts in the reconstructed images) to precision should be made as high as possible. A comparison of tolerance-to-precision ratios for each calibration procedure should indicate which parameters need to be the most rigorously monitored, in order to maximize the system's precision over time.

Precision of the calibration procedures is defined in a number of ways: (1) For the case of geometrical factors such as SCR distance or detector angles, it is determined by the reproducibility of the various scanner movements; (2) for the case of counting factors such as dead-time or beam-hardening, it is determined by the standard deviation (SD) of a number of estimates (done in quick succession) of the particular counting parameter; and (3) for the calibration procedure where output spectra are matched, the

precision is limited by the signal-to-noise ratio in the spectra.

Tolerance levels for each calibration procedure were established by using the two-aluminum-ring phantom illustrated in Figure 36. This phantom was used to simulate actual bone evaluation measurements of the human forearm - or the interior of any cortical bone/trabecular bone structure. Tolerance levels would vary somewhat for different anatomical sites, e.g. tolerance levels for the surrounding tissue of the forearm might be lower. A "region of interest" (ROI) within the larger of the two aluminum rings was analyzed (mean ($\bar{x}$) and SD values of the pixels within the region were calculated) as each of the calibration parameters was separately varied. Depending on the type of artifact induced, either $\bar{x}$ or SD will change predominantly. For example, black and white streaks may occur which might not affect the mean pixel value but would increase SD. Tolerance levels have been arbitrarily defined as a 1% change in either $\bar{x}$ or SD (whichever is dominant). A computer program was used to define the size and location of the ROI within the reconstructed images. In this manner, variation due to different sample sizes (number of pixels) or locations can be eliminated. Tolerance levels were established at two rotational speeds (400 Hz and 1500 Hz) so that the effects of increased noise in the reconstructed images could be evaluated.

The tolerance level for the procedure which matches spectral peaks was determined by varying PMT high voltages for detector beams passing through the ROI. A change in the peak of ± 40 mV was found to produce a $\pm 1\%$ change in $\bar{x}$ for both scan speeds. The precision of the procedure was defined as ± 10 mV (see section 4.1), so a tolerance-to-precision ratio (T/P) of 4 was realized. The tolerance and precision levels of all calibration procedures are shown at the end of this section.

The precision of the fan-beam centering procedure depends on the precision of the horizontal movement ($1.9\mu\text{m}$) and of the position switch ($1.6\mu\text{m}$). Hence precision equals $((1.6)^2 + (1.9)^2)^{1/2} = \pm 2.5\mu\text{m}$. Tolerance was established to be ± 8 motor steps from the center-position or $\pm 25\mu\text{m}$ (for a 1% change in SD) at a scan speed of 400 Hz, and so T/P equals 10. For a scan speed of 1500 Hz, tolerance equaled $\pm 32\mu\text{m}$, so T/P equals 13.

The precision of the SCR distance procedure depends on the precision of the vertical movement ($6.8\mu\text{m}$) and of the position switch ($3.2\mu\text{m}$). Hence precision equals $\pm 7.5\mu\text{m}$; tolerance was determined to be ± 0.95 mm (for a 1% change in $\bar{x}$) at both scan speeds, so T/P equals 125.

The precision of the calibration procedure to determine rotational fan-beam offset depends on: the precision of the rotational movement (0.001°), the position switch (0.01°), and negligible contributions from horizontal and vertical movements. Hence precision equals $\pm 0.01^\circ$, and

tolerance was determined to be $\pm 0.04^\circ$ ($\pm 1\%$ change in SD) at a scan speed of 400 Hz, so T/P equals 4. At 1500 Hz, tolerance equaled $\pm 0.05^\circ$, so T/P equals 5.

The calibration procedure to determine the dead-time correction was repeated several times in quick succession to establish the variation in the coefficients a and b of the 2nd-order polynomial describing the dead-time curve for a particular detector - see section 4.4. The SD or precision was determined to be 1.5% for each parameter. Tolerance ($\pm 1\%$ change in SD) was found to be $\pm 12\%$ (this value represents a $\pm 12\%$ change in the coefficients a and b) for a scan speed of 400 Hz, so T/P equals 8. At a scan speed of 1500 Hz, tolerance equaled $\pm 22\%$, so T/P equals 15.

Beam-hardening values were also determined on several occasions in quick succession and a precision of $\pm 1.5\%$ was calculated. Tolerance ($\pm 1\%$ change in SD) was found to be $\pm 7.5\%$ (this value represents a $\pm 7.5\%$ change in the coefficients describing the beam-hardening curve for a particular detector - see section 4.5) and so T/P equals 5 at a scan speed of 400 Hz. At 1500 Hz, tolerance equaled $\pm 15\%$, so T/P equals 10.

Scans were also performed at various "warm-up" periods of the anode. A tolerance level of about 1 minute warm-up ($\pm 1\%$ change in $\bar{x}$) was determined at a scan speed of 1500 Hz, whereas at 400 Hz, tolerance equaled approximately 2 minutes.

Table 6. Precision and Tolerance of Calibration Methods.

Method	Tolerance (1500 Hz)	Tolerance (400 Hz)	Precision	T/P
Spectrum Peak	± 40 mV	± 40 mV	± 10 mV	4
Centering Fan beam	± 0.032 mm	± 0.025 mm	± 0.0025 mm	13 10
SCR	± 0.95 mm	± 0.95 mm	± 0.75 μ m	125
Detector Angles	$\pm 0.05^\circ$	$\pm 0.04^\circ$	$\pm 0.01^\circ$	5 4
Dead-time	± 22 %	± 12 %	± 1.5 %	15 8
Beam-hardening	± 15 %	± 7.5 %	± 1.5 %	10 5
Warm-up Anode	1 min.	2 min.	-	-

Precision and tolerance levels were not established for the background-count correction because the effect is too small (i.e. background count-rates are insignificant compared to open count-rates).

The tolerance and precision levels for each of the calibration procedures are summarized in Table 6. The tolerance levels are for changes to an individual calibration parameter; combinations of calibration parameter errors would be very difficult to analyze because errors could be both enhanced and diminished; that is, for example, an error in dead-time might increase a pixel value, whereas an error in beam-hardening might decrease the value of that same pixel to give a null effect. Changes over time in these calibration parameters are considered in the next section.

6.4 System Stability

A measure of the system stability is necessary to determine how often the various calibration procedures need to be redone. Electronic drift or a shift in the anode focal spot will affect all of the calibration procedures. For example, a movement of the anode focal spot will change the

intensity spectrum reaching each detector (the intensity spectrum is inhomogeneous - Figure 12); hence all calibration parameters will be affected to some extent.

The system stability can be monitored by running the calibration procedure for matching spectrum peaks and recording changes in PMT high voltages. That is, any change in one of the calibration parameters will cause the intensity spectrum reaching a particular detector to change. Variation in PMT high voltages over time suggests that the system needs to be re-calibrated in this manner on a weekly basis. That is, changes in PMT high voltages over longer periods of time are greater than the precision of the spectrum matching procedure (± 10 mV/Cal Factor [mV/V]).

Unfortunately, the cause of the change in the output spectrum shape cannot be uniquely determined, so a complete re-calibration of the system would be necessary. While it is feasible to re-calibrate the system for the various counting parameters on a weekly basis, the geometric calibration procedures are far more time consuming. Instead, the system is only re-calibrated on a weekly basis for PMT high voltage adjustment. By matching the effective energy of the intensity spectra of all detectors, changes in dead-time, background noise, and beam-hardening effects can be stabilized. for geometric parameters such as SCR distance.

The effects of this modified re-calibration scheme can be monitored by analyzing the drift of the $\bar{x}$ and SD of pixel values in a particular phantom. Using the two-aluminum-ring

phantom again, $\bar{x}$ and SD were established over time in a ROI - the inner 45% region of the larger aluminum ring. Scans were initially performed at speeds of 400 Hz and 1500 Hz in quick succession to establish "normal" variation of $\bar{x}$ and SD values. Scans were then performed over various time intervals to establish the extent of system drift. The results are shown in Figure 38. There is a general decrease in mean pixel value within the ROI over time at both scan speeds; the variation in SD over time is not as pronounced.

A 68% confidence interval for the $\bar{x}$ pixel value in the ROI, at a speed of 400 Hz, is equal to $\pm SD/\sqrt{n} = \pm 10$ pixel units, where SD is the standard deviation (350.12), $\bar{x}$ the pixel value (5312.46), and n the number of pixels (1313) in the ROI. The pixel units form an arbitrary numbering scale and need to be standardized to a material of known density before they have any physical meaning (n.b. beam-hardening and dead-time corrections were all relative; the slope m was arbitrarily set to some value, therefore the pixel value units are all relative and in themselves have no physical meaning). The relative value $(\pm 10/\bar{x}) \cdot 100\% = \pm 0.19\%$ constitutes the precision of estimating the mean pixel value in the ROI (based on a 68% confidence interval).

At a scan speed of 1500 Hz, $\bar{x} = 5304.33$ pixel value units and SD = 634.91. A confidence interval of ± 18 pixel units is realized to give a precision of estimating $\bar{x}$ in the ROI of $\pm 0.34\%$ (at a 68% confidence level). Hence by increasing the number of pixels within the ROI and by

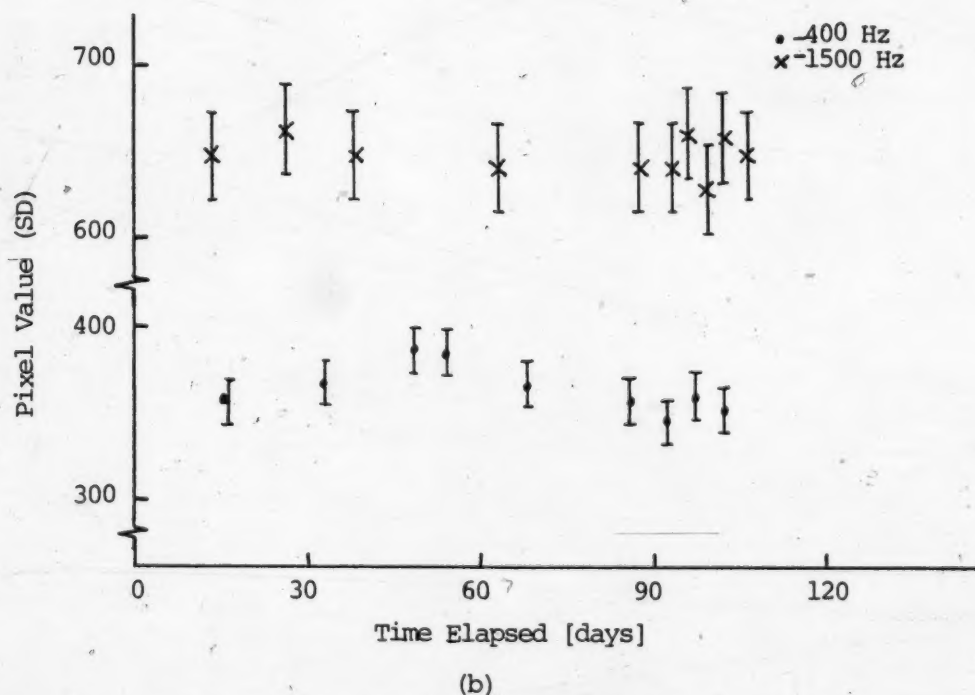
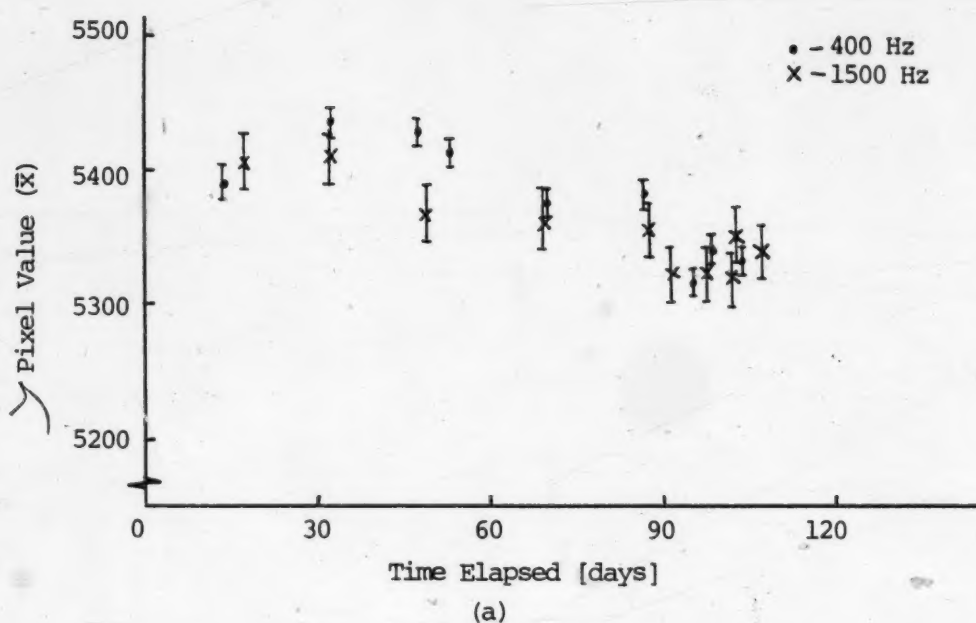


Figure 38. Stability of the CT System.

- (a) Variation of $\bar{x}$ pixel value in ROI over time,
 (b) variation in SD of pixel values in ROI over time.

decreasing the SD in the pixel units (i.e. recording more counts per projection value) the precision of estimating $\bar{x}$ can be improved.

To maximize the precision of the bone evaluation method (i.e. an estimation of $\bar{x}$ over time) the pixel values should be regularly standardized against a phantom of known density (e.g. a plexiglass phantom filled with water, or a mineral concentration of K_2HPO_4). In this manner, any drifting of the various calibration parameters (not corrected in the weekly spectrum matching re-calibration procedure) will be accounted for during this standardization. The regularity of this standardization will depend on the number of counts recorded per projection value. For example, the precision of estimating $\bar{x}$ at a scan speed of 400 Hz is approximately twice that of the 1500 Hz scan, hence the standardization procedure would need to be performed twice as often.

6.5 Patient Measurements

To illustrate the clinical potential of the CT scanner, a series of measurements were made on a cadaver specimen. Distal and proximal regions of a wrist of a female cadaver were scanned (Figure 39a,b). The convoluted surface of the soft-tissue is due to the age of the cadaver specimen (about 1 year). In addition, the specimen was dissected prior to the scan which explains the "broken" appearance of the soft tissue. The high spatial resolution of this scanner provides very accurate bone edge definition, thus making it easier to

reproducibly define the trabecular bone region of a patient's wrist. In addition, the thigh of a female cadaver was scanned to illustrate the range of object sizes which can be measured (Figure 39c). Each scan took approximately 2.0 minutes (1500 Hz) and was measured at 50.0 kVp, 2.8 mA, and with a 0.70 mm copper filter.

6.6 Conclusions

A number of calibration procedures have been implemented which (1) define various geometric parameters such as SCR distance, fan beam centering, and detector angles, and (2) determine various counting parameters such as spectrum matching, "background", dead-time, and beam-hardening corrections. The high T/P ratios indicate that all calibration parameters can be adequately defined. The procedures which need to be monitored the most rigorously (i.e. lowest T/P ratios) are the spectrum matching, detector angles, and beam-hardening corrections. Tolerance levels and T/P ratios generally increase as scan speeds are increased (where all other scan parameters remain constant). That is, statistical noise in the reconstructed images conceals calibration errors to a greater extent as photon counts per projection value decrease. However, scanning procedures should not be designed solely for high T/P ratios; the most important criterion for this CT scanner is high precision (for bone density evaluation), thus slower scan speeds (with their better counting statistics) must be

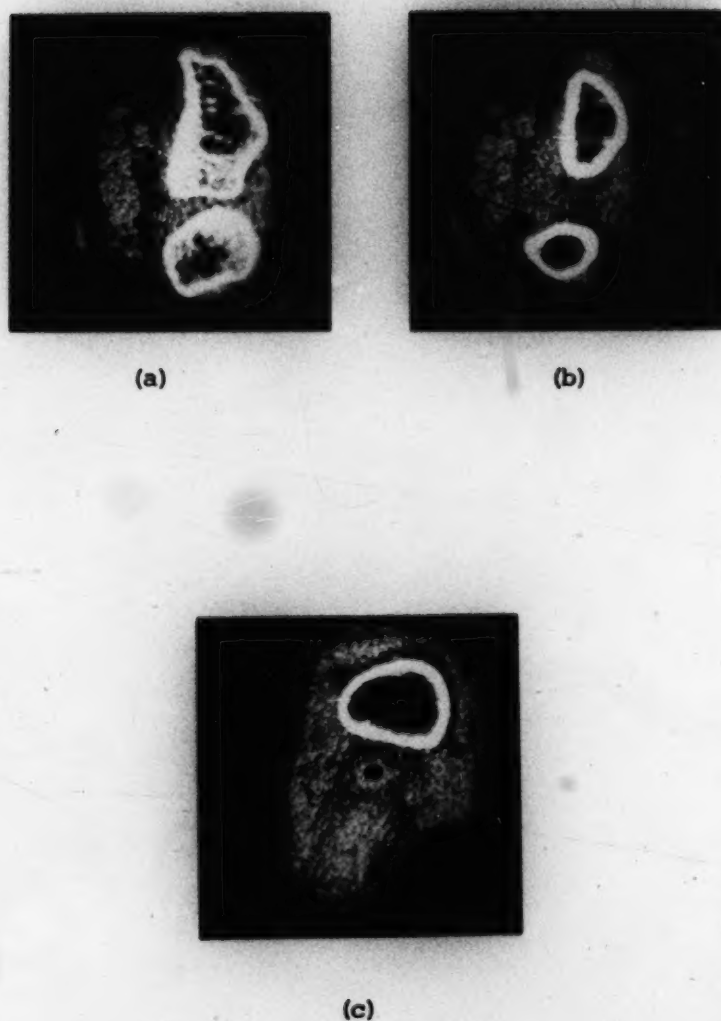


Figure 39. Reconstructed Images of Human Forearm and Thigh.
(a) Distal view of forearm, i.e. towards the wrist;
(b) proximal view, i.e. towards the elbow; (c) human thigh.

used.

The system needs to be re-calibrated periodically due to electronic drift or tube anode wear. System stability can be monitored by observing any changes in detector output spectra. Unfortunately, the cause of the change in the output spectrum shape cannot be uniquely deduced. To correct for changes in spectrum shape, the spectrum matching procedure is performed regularly (on a weekly basis). This calibration updating can correct for changes in dead-time, beam-hardening, and spectrum peak positioning; however, it does not correct for changes in any of the geometric parameters.

To determine how well this partial re-calibration process works, the mean pixel values and the SD of a region of an aluminum-plexiglass phantom were recorded over time. By analyzing pixel values within the inner 45% region of an aluminum ring, the bone evaluation procedure for patients was simulated. The drift in $\bar{x}$ over a three month period was found to be about three times the "normal" variation, whereas SD showed little change in this same time interval. Hence in order to maximize the CT system's precision (for bone evaluation) a standardization of the pixel values should be made (using a plexiglass-water phantom, for example) on a regular basis.

7. Future Developments

The system at present can only be used for bone density measurements; however, by increasing the X-ray flux, some soft-tissue (e.g. fat) discrimination should be possible. One application could involve pediatric brain scanning. In order to see small density changes, high numbers of photons must be recorded to reduce statistical counting noise; either count-rates or measurement time must be increased. Currently the maximum count-rate is about 0.6 MHz, and typical measurement times are between 2 and 4.5 minutes (1500 Hz and 400 Hz rotational scan speeds, respectively). A longer measurement time increases the probability of patient movement and so is undesirable. The X-ray flux can be increased by either decreasing the thickness of the copper filter or by increasing the collimator slot size. The latter causes problems due to scattering and also decreases spatial resolution by increasing the beam width. Utilizing a thinner copper pre-filter changes the shape of the X-ray spectrum, but the counting parameters, such as dead-time, background counts, and beam-hardening corrections, of the system can easily be re-calibrated.

7.1 Pulse Amplification

The main problem associated with higher count-rates is that the counting system will become paralyzed, that is the voltage level produced by the electron pulses from the PMT and pre-amplifier will remain above the threshold voltage

level of the discriminator modules, hence no pulses will be recorded.

The mode of operation of the discriminator units cannot be changed; the problem lies with the performance characteristics of the LeCroy TRA1000 pre-amplifier. The output voltage of the TRA1000 decreases significantly as frequency response is increased [19], hence a compromise between pulse height and frequency response had to be reached.

In the near future, the single stage TRA1000 pre-amplifiers will be replaced with a two stage system where the first stage is primarily used as a current-to-voltage converter and the second stage is used for signal amplification. In this manner, little compromise is necessary between output voltage and frequency response. This new amplifier system should handle count-rates of up to 1.2 MHz, and for typical bone density scans, measurement times could be decreased to about 30 seconds.

7.2 Adjustable Counter-Weight

In order to ensure the possibility of fast scan times, a static balance of the rotating scanner parts needs to be established. That is, the weight of the detector assembly must be counter-balanced at all SCR distances. One method would be to use a counter-weight attached to a ball screw with the same pitch as that of the middle plate. In this way, any movement of the middle motor (the motor which

adjusts SCR distance) causes the counter-weight to move in the opposite direction so that the static balance is maintained. Consequently, the torque on the rotational motor shaft is minimized at all SCR distances and maximum rotational speed is possible.

7.3 Re-Packaging of Radiation Detectors

Another possible means of increasing scan speeds would be to use a higher packaging density of radiation detectors. Currently, the CT system contains 26 NaI(Tl) crystal-PMT detectors; increasing the number of detectors to 64 or 128 (by stacking the detectors in 3 or more planes and using light guides between crystals and PMTs) would reduce the number of rotations per scan. Hence, scan times could be reduced to less than 15 seconds, and the dose per scan would also be reduced - due to more efficient data collection.

7.4 CAMAC Controller/Microcomputer

An IS-11-microcomputer/CAMAC-crate-controller will eventually replace the 3912-Unibus-crate-controller currently being used. The primary function of the microcomputer of this unit will be to free up the multi-user VAX system. For example, the microcomputer's 256 Kbytes of RAM allows data collected after each scaler-interrupt pulse to be transferred to memory in the CAMAC crate instead of to a temporary buffer in the VAX. With the present system, 541 data transfers between CAMAC and the VAX occur for every

scanner rotation whereas with the IS-11 system only one data transfer per rotation would be needed.

In addition, data collection for one rotation would no longer be restricted to 64 Kbytes: the data for one rotation could be transferred in 64 Kbyte blocks from the CAMAC microcomputer to a file in the VAX.

7.5 Dual Energy CT

Single energy quantitative CT (SECT) measurements have poor accuracy for determining bone mineral density (BMD). In order for accurate density values to be determined, the object scanned can only consist of 2 components. However, for patient measurements, three components are involved, namely bone, marrow, and fat. Generally, SECT procedures assume only the bone and marrow components. By assuming that there is no fat content, the bone density measurements provide underestimated values.

A method to produce more accurate BMD values, at a cost of poorer precision, is dual energy CT (DECT). By scanning an object at two energies for which the linear attenuation coefficients vary significantly, enough information is gathered to solve the density values of a three component system.

A CT unit specifically designed for DECT has the ability of rapid kilovoltage switching so that only one scan is needed to produce images at two energies. This new specialized CT scanner however, has no means of rapid

kilovoltage switching so two scans would be necessary, one at each energy. An advantage of this system is that the new scanner would operate at 50 kVp and 70 kVp. Large changes in attenuation coefficient values will then be possible because at 50 kVp, the photoelectric interaction dominates, whereas at 70 kVp, the Compton interaction dominates.

These are a few of the future developments which will make this specialized CT scanner capable of providing accurate and precise BMD values. The scanner will have the added ability to be used for soft tissue discrimination with spatial resolution capabilities equal to any commercial whole body CT scanner.

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Appendix I

Bone is a dynamic organ consisting of collagen fibers which provide tensile strength and of hydroxyapatite crystals containing calcium and phosphate which provide compressional strength. Osteoblast cells and osteoclast cells along with various activators form the basic bone remodeling units (BRUs) which dictate the rate of turnover of bone in adults. Some kind of stimulus activates a BRU, whereupon a phase of osteoclast bone resorption occurs for about 20 days. The osteoclasts then disappear and are replaced by osteoblasts which form new bone in the cavity over the next 70-90 days. There are approximately 10⁶ active BRUs in a normal skeleton at any one time, and each BRU turns over a packet of bone of about 0.1 mm³ volume. Therefore a balance between osteoclast activity (bone resorption) and osteoblast activity (bone formation) is necessary to maintain bone mass.

In adults over the age of 35, osteoclast activity is slightly greater than osteoblast activity, causing a drop in bone density of about 1-2% per year. That is, at a macroscopic level the bone becomes more porous with age. When bone density decreases to the point where there is an onset of skeletal biomechanical incompetence (e.g. hip, wrist, or spinal fracture) the bone is considered osteoporotic. This critical level is generally thought to be about 80 mg/ml K₂HPO₄ eq [44,45]. The disease osteoporosis affects primarily post-menopausal women. It is believed that the drop in estrogen levels following menopause causes rapid

bone loss for about 3 years, after which time bone loss stabilizes again to 1-2% per year. However, the damage is now done and approximately one in every four women will eventually suffer from osteoporosis. It is estimated that hospital care for osteoporotic patients costs about \$5 billion a year in the U.S.A. and is expected to top \$20 billion by the end of the century as an increasing percentage of the population reaches old age. Billions of dollars to the health care system can be saved if bone density in the elderly can be kept above the critical level for fractures.

A first approach to combating bone density loss would be to find a pharmaceutical agent which either depresses osteoclast activity or enhances osteoblast activity, or both. Despite enormous research efforts, no drug has appeared which depresses only resorption or stimulates only formation. All evidence suggests that osteoblasts and osteoclasts are functionally dependent populations of skeletal cells. Furthermore, BRU populations in any given region of bone are found in all different stages of their lifetimes, that is they display the property of incoherence in time. Hence, applying a depressant to the bone simply depresses all stages of BRU activity causing a null effect.

The latest concept for combating osteoporosis is the ADFR - activation, depression, free, and repeat - method (Figure 40) [46]. A large number of BRUs (10^4) are simultaneously activated at time t , by agents such as PTH,

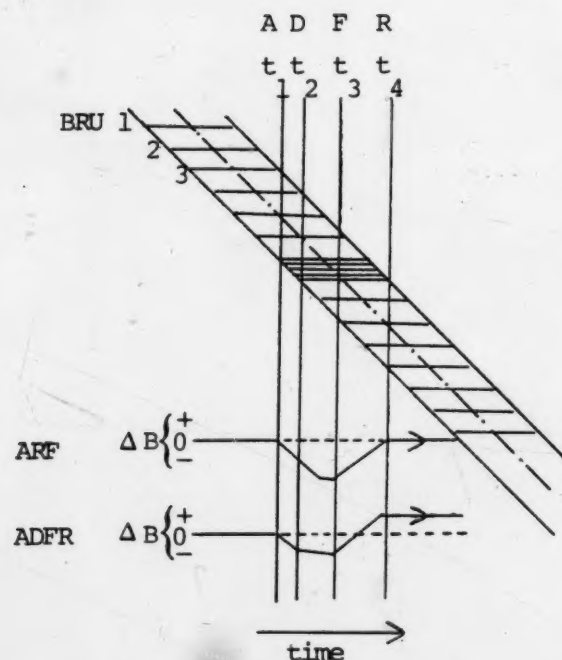


Figure 40. Temporal Sequence of ADFR Method.

BRU activity normally exhibits temporal incoherence, however applying a stimulus causes many BRUs to be activated over a short period of time thus eliminating temporal incoherence. The normal BRU sequence (ARF) produces no net change in ΔB , the bone mass per unit volume. The ADFR method produces a net increase in ΔB .

growth hormone, or vitamin D metabolites. This in effect eliminates the temporal incoherence problems of normal BRU activity. At time t_2 , a depressor such as calcitonin is administered causing a net effect of depressed osteoclast activity. At time t_3 , when osteoblast activity dominates, the depressant is removed and remodelling activity is returned to normal. That is, upon BRU activation at t_1 , a 'pre-programmed' number of osteoblast cells will be activated at t_3 , regardless of the number of active osteoclast cells. The net effect is: bone deposition is

greater than bone resorption, or ΔB , the bone mass per unit volume, is increased. The process is repeated to increase bone density to normal levels.

In order to monitor the effects that various reagents have on the ADFR process, bone density must be measured with high precision in longitudinal studies. Computed tomography is the preferred method for these bone density measurements.

Appendix II

Geometrical and Counting Parameters of Scanner

Detector'	PMTHV (Volts)	Open-Count (cps)	Background (cps)	Angle (θ)	Cal Factor (mV/V)
1	913	75169	1.24	24.9521	3.3
2	944	83613	1.12	22.9519	4.1
3	844	126732	1.18	20.9523	3.8
4	979	67511	1.29	18.9535	3.1
5	938	70725	1.39	16.9554	3.3
6	956	80245	1.33	14.9580	3.5
7	888	113391	1.21	12.9613	4.7
8	929	96647	1.27	10.9654	3.8
9	792	211003	1.04	8.9701	4.3
10	798	211586	0.97	6.9755	4.3
11	888	197936	1.22	4.9817	4.3
12	777	211688	1.07	2.9885	4.3
13	826	206824	1.01	0.9959	3.8
14	810	213297	1.09	-0.9959	4.1
15	838	144785	1.24	-2.9871	4.3
16	791	228840	1.09	-4.9777	5.2
17	794	199396	1.05	-6.9677	5.3
18	802	206690	1.05	-8.9570	5.0
19	882	104712	1.11	-10.9457	4.8
20	898	101062	1.32	-12.9338	4.1
21	911	98642	1.21	-14.9214	3.6
22	971	77061	1.35	-16.9084	3.6
23	882	89593	1.32	-18.8949	3.1
24	914	97588	1.54	-20.8808	3.8
25	879	96431	1.25	-22.8662	3.9
26	932	59715	1.39	-24.8512	3.1

'Detectors 1-8 and 19-26 are the "old" detector version, whereas detectors 9-18 are the "new" version. This is why there are large differences in open-counts, and PMT high voltages.

Appendix III

μP EPROM (0-1000)

M.A.	Function
0000	Interrupt vector 0 (WP,PC): Reset μP
0004	1: Not used
0008	2: Status
000C	3: Stop motors
0010	4: Not used
0014	6: Read message from VAX
0018	6: Write message to VAX
001C	7: Axial motor (MOT1)
0020	8: Rotational motor (MOT2)
0024	9: Middle motor (MOT3)
0028	10: Front motor (MOT4)
002C	11: MOT1 position switch
0030	12: MOT2 position switch
0034	13: MOT3 position switch
0038	14: MOT4 position switch
003C	15: Debounce
0048	Subroutine to initialize a read operation from the TM990/310 (I/O) board
0092	Subroutine to initialize a write operation from the TM990/310 board
00F4	Subroutine to obtain status of current operation (0=clear, 1=read, 2=write)
00AE	Subroutine to read message from VAX
0176	Subroutine to check 0000, FFFF, and CHECKSUM on incoming message
01A8	Subroutine to prompt VAX to repeat message up to 3 times. If error persists emergency stop routine is called.
01B8	O.K. message to VAX: 0 0 0
01E2	Error message to VAX: F 0 0
020C	Status message to VAX: 1 STATUS 1
0224	Continue message to VAX: 3 0 0
0252	Position message to VAX: 4 MOTOR# POS
0276	Early start motor message: 5 MOTOR# 0
02BC	Subroutine to write message to VAX
02F4	Subroutine to send status message to VAX
0368	Subroutine to stop motors in emergency as a result of lost communication with VAX
0700	Subroutine to enable position switches
0714	Subroutine to initialize MOT1 forward movement
071E	MOT1 backward
0744	MOT2 CW
074E	MOT2 CCW
0776	MOT3 up
0780	MOT3 down
07A6	MOT4 left
07B0	MOT4 right

07D6	Subroutine to check state of motors. When all motors have stopped, an end of program message is sent. Early start condition for a motor is also monitored.
08F4	Subroutine to place MOTOR#, direction, and motion description table into appropriate workspace.
0A64	Subroutine for motor interrupt service: generates motor pulses, decrements SINT, STPCNT, and STSNM; sends pulse to scalars.
0B70	Emergency stop subroutine: motors are stopped in the order 1, 2, 3, and 4.

 μ P RAM (F000-FFFF)

M.A.	Function
F000	Transfer address of motion description table for MOT1
F25E	MOT2
F4B6	MOT3
F70E	MOT4
F966	Location for communication time-out
F968	Location for motor flags
FFE8	Location for Data read from CAMAC
FFF0	Location for Data write to CAMAC
FFFA	Location for storing 0000 from VAX
FFFC	Location for FFFF from VAX
FFFE	Location for CHECKSUM from VAX
FEC0	Timer workspace
FEE0	Main workspace
FF00	Interrupt workspace
FF20	Upload workspace
FF40	MOT1 workspace
FF6A	MOT2 workspace
FF94	MOT3 workspace
FFBE	MOT4 workspace

Workspace Registers of MOT1 to MOT4

Register	Contents
0	Used for any calculations
1	Pointer for each word in movement description file
2	Motor bit number with direction information
3	Direction information (CW or CCW)
4	SINT
5	STSNM
6	Motor base - contains information of which motor to start
7	STPCNT
8	Workspace address
9	Position information
10	Used for comparisons
11	Branch link for return from subroutine
12	CRU - base address for I/O
13	Stores old WP
14	Stores old PC
15	Stores old status register
16	STSNM - for reloading purposes
17	End-of-motion flag
18	Motor base - for reloading purposes
19	Timer offset - for reloading purposes
20	Re-entry flag - set to logic 0 after a subroutine is performed